Hearing Aid

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ISBN 978-93-5419-433-7

Preface

This book fills up an important gap that existed in the post graduate learning resource. Otolaryngologist had a poor insight as far as this topic is concerned. The author feels that all practicing otolaryngologist should have a basic understanding of hearing aid, how it functions, and how to prescribe them. More than anything else they should also be aware about the limitations of the hearing aid.

This book should enlighten the students of otolaryngology on this vital topic. This topic is a little bit boring because of the physics involved. The author wishes to allay the fears of the student that prior knowledge of acoustic and electro-physics is not needed. All the basic stuff have been explained on a need to know basis with clear illustrations.

Standard text books in otolaryngology don't cover this topic with clarity leaving the otolaryngologist to fend for themselves when it comes to hearing aids.

The author feels a practicing otologist should have a clear understanding of the concept of hearing aids and the development that have been taking place in this field. Clear idea of this topic will ensure that the otologist to whom the patient first lands up give proper guidance to the patient. Otologist should also be in a position to offer alternate non invasive options of improving the patient's hearing. The author is of the opinion that proper motivation of the patient to use hearing aids will help them to avoid risky surgical procedures to improve hearing in otosclerosis.

As far as sensorineural hearing loss hearing aid happens to be the only viable solution that is available to the patient. The author also wishes to stress on the importance of using bilateral hearing aids in patients with bilateral hearing loss (be it conductive or sensorineural hearing loss).

This book attempts to impart vital knowledge on this topic using below upwards concept, making it easy for a novice to understand the topic. The author hopes that this book will be useful to the students of otolaryngology as well as the practitioners of otolaryngology.

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Introduction

Hearing disability happens to be one of the most common disability in human population and presents a challenge to the individual in speech recognition, communication and language acquisition. In daily life speech is often heard among a variety of sounds and noisy backgrounds which makes communication even more challenging. It has been proven the processing of ongoing auditory streams increases the cognitive load imposed by the listening task. This increase in the cognitive load by the listening task causes increased levels of mental stress and fatigue, lack of energy and stress related sickness. These patients tend to be withdrawn from leisure and social roles.

It has been estimated that hearing loss affects nearly 250 million people in the world. Humans are highly dependent on their senses. The sense organs play an important role in shaping both physical and psychological growth and behavior. All sense organs play an important part in shaping both physical and psychological growth and behavior. Hearing and vision happen to be distant senses and are the most crucial ones.

Hearing aids are typically used to correct the loss of audibility introduced by hearing impairment. Modern hearing aids provide a range of signal processing algorithms. These include amplitude compression, directional microphones and noise reduction. The basic purpose of these algorithms is to improve speech intelligibility and listening comfort. If hearing impairment increases the listening effort, then it is essential to investigate whether hearing aids can reverse this aspect of hearing loss. The most prevalent degree of hearing loss ranges between mild-to-moderate affecting more than 90% of all adults with hearing loss. This figure is based on an UK study. Sensitivity of hearing can be assessed according to pure-tone thresholds across five different octave frequencies (0.25, 0.5, 1, 2, 4, kHz).

Hearing level in mild hearing loss = 20-40 dB

Hearing level in moderate hearing loss = 41-70 dB in the better hearing ear.

Audiological rehabilitation is efficacious when it reduces communication difficulties, enhances psychosocial well-being. It should be stressed that even though hearing aids constitute the single most important part of auditory rehabilitation process, they constitute only one part and may not be even indicated for certain hearing-impaired individuals. Hearing aids are not taken up by nearly 60% of these patients due to so many reasons, the most common happens to be social taboo. Despite the fact that hearing deterioration gradually increases from the age of 55, hearing aids are not typically adopted till they reach the age of 70. It should be stressed that hearing aid fitting from earlier age can result in substantial benefits. Conventional hearing aids do not restore hearing but makes sounds more audible through electro-acoustic amplification. Hearing aids are regulated medical devices that deliver sound into the ear canal via air / bone conduction. Bone conduction hearing aids / Bone anchored hearing aids are indicated for patients with external auditory canal atresia which could make wearing hearing aids rather difficult to comply.

Depending on the type of technology used hearing aids can be classified into:

Analog

Digital

Depending on the model they can further be sub-clas-

sified into:

Body worn hearing aid
Behind the ear hearing aid

In the ear hearing aid
Receiver in the canal hearing aid

Physical characteristics of sound waves that should be understood before deciding on use of hearing aids include:

Frequency:
This is the rate at which the sound fluctuates.

Pitch:
This is the time taken for a repetitive fluctuation to repeat.
Wavelength:
This is the distance over which its wave form repeats.
Diffraction:
This is the way the sound bends around obstacles
Pressure / Sound pressure level:
This is the strength of the sound wave
Spectrum:
This is a breakup of a complex sound into pure tone components at different frequencies.

Amplifiers present inside the hearing aids can be classified into linear and nonlinear. For sounds of a given frequency, linear amplifiers amplify by the same amount regardless of the level of the signal, or what other sounds are simultaneously present. Whereas nonlinear amplifiers vary with the amplitude of the signal input to the amplifier. The degree of amplification can be represented as a graph of gain versus frequency (gain: frequency response), or as a graph of output versus input level (I-O curve). The highest sound level produced by the hearing aid is known as the saturation sound pressure level (SSPL). SSPL is usually estimated by measuring the output sound pressure level for 90dB SPL input (OSPL90).

Patients with sensorineural hearing loss have several deficits to overcome. Some sounds could be inaudible, other sounds can be detected because part of the spectra is audible, not identifiable because other parts of their spectra remain inaudible. The range of levels between the weakest sound that can be heard and the most intense sound that can be tolerated is less for a person with sensorineural hearing loss than for a normal hearing person. Ideal hearing aid should amplify weak sounds more than they amplify intense sounds. In addition, sensorineural impairment diminishes the ability of a person to detect and analyze energy at one frequency in the presence of energy at other frequencies. These patients also have difficulty in hearing a signal that rapidly follows or is rapidly followed by a different signal. These problems could mean that noise, or even other parts of the speech spectrum, will mask speech more than would be the case for a normal hearing person.

Sensorineural hearing impairment needs a signal to noise ratio greater than normal in order to communicate effectively, even when sounds have been amplified by a hearing aid. In conductive hearing loss there is simple attenuation of sound as it passes through the middle ear, so amplification provided by hearing aids helps in restoring hearing to normal levels.

The size of the hearing aid has shown a constant trend of decreasing. The currently available hearing aids are so small they can fit fully into the external auditory canal making it nearly not visible to others.

Problems faced by patients with sensorineural hearing loss:

1. Some of the sounds could be inaudible.

2. Some sounds could be detected but would not be correctly interpreted because part of their spectra could be inaudible. This is more so for the high frequency spectrum.

3. The range of sound level between the weakest sound that can be heard and the most intense sound that can be tolerated is less for a person with sensorineural hearing loss when compared to normal individuals. In order to compensate for this effect, hearing aids will have to amplify weak sounds more than that of intense sounds.

4. Sensorineural hearing loss impairs the ability of a person to detect and analyze energy at one frequency in the presence of energy at other frequencies.

5. In these patients there is also a decreased ability to detect and analyze energy at one frequency when it is rapidly followed by another signal.

6. Hearing impaired persons find problems in separating sounds on the basis of the direction from which they arrive.

The decreased (frequency, temporal and spatial) resolution would cause a noise or even other parts of the speech spectrum to mask speech much more than normal individuals.
Patients with sensorineural hearing loss will need a signal to noise ratio greater than that of normal hearing individuals to comprehend spoken voice. This is true even when sounds are amplified by hearing aid. This should be contrasted with that of patients with conductive hearing loss where there is attenuation of sound as it traverses the middle ear cavity. Amplification provided by hearing aids in these patients restores hearing to a great extent.

Causes of sensorineural hearing loss:

Loss of inner hair cell function

Loss of outer hair cell function

Reduction in the cochlear electrical potential

Changes in the mechanical properties of cochlea

Types of Amplifiers that are present in the hearing aid:

Amplifiers inside hearing aids can be classified as linear or non linear.

Linear amplifiers - Amplifies the sound by the same amount regardless of the frequency. It also amplifies the background noise also.

This is an electronic circuit whose output is proportional to its input and is capable of delivering more power into a load. Not all linear amplifiers are equal and circuits that go into their design provides trade offs. So there is nothing like 100% amplification of input sound and they are classified into three classes based on their amplification ability.

Class - A amplifiers are very inefficient and their efficiency in amplification does not exceed 50%. The semiconductor / vacuum tube in this class of amplifiers conduct throughout the entire RF cycle.

Class - B amplifiers can at the most be 60-65% efficient. The semiconductor in this device conducts through half the cycle but needs lots of power to run.

Class - C amplifiers are more efficient and can be about 75% efficient in amplifying sounds.

Non linear amplifiers - Amplification provided by nonlinear amplifiers varies with the amplitude of the signal input into the amplifier.

The amount of sound amplification can be represented graphically as a graph of gain versus frequency (gain-frequency response) of as a graph of output level versus input level (I-O curve).

The highest level of amplification provided by hearing aid is known as saturation sound pressure level (SSPL). This value is usually estimated by measuring the output sound pressure level for a 90 dB SPL input.

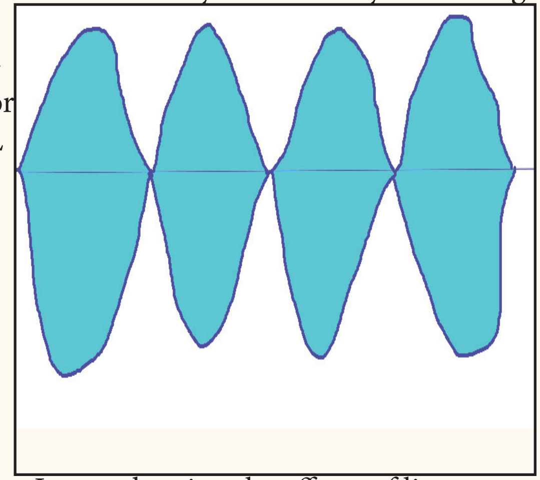


Image showing the effects of linear amplification

Non linear amplifiers - Amplification provided by nonlinear amplifier varies with the amplitude of the signal input to the amplifier. Only the weak signals are amplified in this set up. Non linear amplification is widely used in hearing aids for the following reasons:

1. Already loud sounds will not be amplified (peak clipping effect). In fact the amplitude of loud sounds a clipped thereby preventing those sounds from excessively stimulating the cochlear hair cells.

2. Cochlear hearing loss are usually non linear in nature. In this type of hearing loss sensitivity for weak sounds are impaired whereas for loud sounds it is unaffected. This reduces the auditory dynamic range for certain frequency regions.

Automatic Gain Control:

The difficulties associated with decreased dynamic range can be reduced by the use of automatic gain control (AGC) according to Steinberg & Gardner (1937). Automatic gain control system works on the basis of signal levels. It amplifies weak sounds more than the stronger ones, resulting in the wide dynamic range of the input signal being compressed into a smaller dynamic range at the output. Hence all the AGC systems are known as "compressors".

AGC reduces the volume if the signal is strong and raises it when it is weak. For this to happen the signal should pass through a detector stage before reaching the amplifier. Only the signal that needs to be amplified goes to the amplifier containing a diode and capacitor.

Automatic Volume Control:

Automatic volume control systems are intended to adjust the gain automatically for different listening situations to relieve the user of the need to adjust the volume control manually. If the recovery time of the AGC circuit is set greater than a few hundred milliseconds then the gain changes slowly with changes in the sound level.

Automatic volume controls provide more amplification for soft voices when listening conditions are quiet and less overall amplification when sounds and conditions are loud.

Many newer hearing aids now have:

Automatic volume controls that provide more amplification for soft voices when the listening conditions are quiet and less overall amplification when sounds and conditions are loud.
Directional microphones that improve the ability of the hearing aid user to understand conversation in background noise by reducing amplification for sounds coming from behind the listener and maintaining amplification of speech coming from the front. Some hearing aids can also be adjusted to reduce amplification for sounds coming from one side and maintaining amplification for speech and sounds coming the opposite side; this would be helpful in a situation such as traveling in a car.
Noise reduction circuitry that improves the user's listening comfort in louder environments by reducing amplification for pitches where constant noise is detected.
Automatic phone programs that "turn on" when the phone receiver is near and "turn off" when hanging up and moving away from the phone.
Circuits that recognize music and adjust the hearing aid settings for improved music sound quality.
Feedback cancellation circuits that monitor for hearing aid "whistling" and reduces, eliminates or prevents it.
Impact-noise sensors that reduce amplification for sudden loud sounds such as a door slam or a shout.
Wind-noise sensors that identify the presence of wind and actively reduce the loudness and annoyance of that signal.

Decreased audibility

Hearing impaired persons don't hear some sounds at all. Individuals with severe / profound sensorineural hearing loss may not hear speech sounds at all. For them to hear speech sounds they should be shouted at a close range. Persons with mild to moderate hearing loss could hear some sounds but not the others. Classically softer phonemes (usually the consonants) may not be heard at all.

These patients also have trouble in understanding speech because essential parts of some phonemes are not audible at all. In order to recognize speech sounds, the auditory system needs to determine which fequencies contain the most energy.

The vowel oo is differentiated from the vowel ee by the location of the second most intense region (i.e. the second formant). If the hearing loss involves the frequency where the second most intense region is present then both these vowels would sound identical to the patient. Both these sounds are heard but cannot be differentiated into two different sounds (Fig 1).

High frequency components of speech are weaker than that of low frequency components. In 90% of adults and in majority of children the degree of hearing loss extends from 500 Hz to 4 KHz. Commonly hearing impaired people would miss out on the high frequency information. It must also be stressed that the loudness of the speech generally originates from the low frequency components. These patients will hence say that the speech is loud enough, but is not clear.

In order to overcome this problem the hearing aid should amplify the weakest component of the speech and also the the frequencies at which the hearing loss is the greatest.

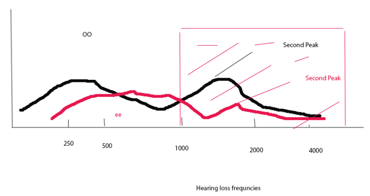


Image showing two peaks for two vowel sounds i.e. OO (Black curve) and EE (Red curve). Red box indicates the frequencies involved in hearing loss. It becomes difficult to hear the second peak and hence these two vowels will not be differentiated.

Figure 1 showing two peaks of vowels OO and EE.

The selection of appropriate hearing aid should involve adjusting the amount of gain provided at each frequency.

Decreased Dynamic Range

Soft / weak sounds can be made audible simply by amplifying them. It should be pointed out that it is not appropriate to amplify all sounds by the same amount that is used to amplify the soft sounds. Sensorineural hearing loss increases the threshold of hearing much more than it increases the threshold of loudness discomfort. In patients with mild / moderate hearing losses there is likely to be very little increase in loudness discomfort level, even though the threshold for hearing has increased by a value up to 50 dB.

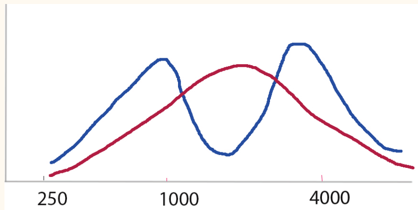
The dynamic range of the ear with sensorineural hearing loss (the amount by which the discomfort threshold exceeds the threshold of audibility) will be less than that of the normal hearing ear.

Dynamic range compression is a concept in audio signal processing that reduces the volume of loud sounds / amplifies quite sounds thereby reducing or compression the audio signal's dynamic range.

Decreased frequency resolution

Patients with sensorineural hearing loss have difficulty in separating sounds of different frequencies. It should be noted the different frequencies are sensed by different regions of cochlea. In normal persons a narrow band sound (sound which concentrates power within a restricted range of frequencies) produces a clearly defined region of relatively strong vibration centered at one position of basement membrane, causing a clearly defined region of activity within the auditory cortex.

In the presence of background noise that contains some energy at the frequency close to one of the components of speech sound, normal ear can do a great job of sending seperate signals to the brain, one for each region of intense activity in the cochlea. (Fig 1.1) The brain can consider all the spatial information that it is getting, as well as visual information about the context of the message (if it is speech) and could ignore the background noise. The ear has frequency resolution that is precise to enable brain to separate speech from noise, provided the speech component and the noise are sufficiently separated in frequency.



Blue cure showing two peaks the first one belongs to noise spectrum and the second peak belongs to the speech spectrum. (Normal hearing).

Red curve which has merged the two peaks as they cannot be differentiated this happens in a deaf ear.

Figure 1.1 showing the response of normal ear when speech is heard along with noise (blue curve) and the response of a deaf ear (red curve) to the same.

Person with sensorineural hearing loss has decreased frequency resolution. The outer hair cells of the cochlea are known to increase the sensitivity of the cochlea for frequencies to which the corresponding part of the cochlea is tuned. Loss of outer hair cells will cause that portion of the cochlea to lose their amplifying ability, and the cochlea hence loses some of its frequency selectivity. Psycho acoustically, this shows up as flatter masking curves and tuning

curves. This deficit causes a single broad region of activity in the cochlea rather than two separate regions (in the presence of noise and voice). This causes difficulty for the brain as it is unable to untangle the signal from that of noise.

The degree of reduced frequency selectivity, and its impact on understanding speech increases with the degree of hearing loss. It is also pointed out that even normal hearing people have poorer resolution at high intensity levels than at lower levels. People with hearing impairment (that too those with profound hearing loss) will have to listen at high levels if they are to achieve sufficient levels of audibility. They hence face difficulty in separating sounds, due to the fact that the cochlea is damaged and partly due to the need to listen the sound at elevated levels.

If the speech and noise are in the same frequency region and they happen to arrive from the same direction and get mixed together inside the hearing aid, there is no way the hearing aid could separate these two sounds to enhance intelligibility. The currently available hearing aids can only minimize the problems caused by decreased frequency resolution. This is usually done by:

1. Keeping noise out of the hearing aid by picking up the signal remotely and transmitting it to the hearing aid.

2. Using a directional microphone to emphasize wanted sounds coming from one direction / or partially suppressing unwanted sounds coming from other directions. This is commonly used in hearing aids where the sounds coming from the front of the person wearing the aid is emphasized more than that coming from other directions. This increases speech intelligibility quotient.

3. By providing an appropriate variation of gain with frequency so that the low frequency parts of speech or noise do not mask the high frequency parts of speech. Hence frequency regions dominated by noise are not louder than frequency regions dominated by speech.

Decreased temporal resolution:

This is a very general term. Intense sounds can mask weaker sounds that immediately precedes them or immediately follow them. This is known as temporal masking in lay terms. This tends to occur more commonly in patients with sensorineural hearing loss than those with normal hearing thereby affecting intelligibility of speech. The increased temporal masking is caused by impaired cochlea which is unable to increase its sensitivity after the masking sound ceases as it happens in a normal cochlea.

In real-life scenario noise tends to fluctuate rapidly, and normal hearing individuals can extract useful bits and pieces of information during the weaker moments of the background noise. This is known as listening in the gaps. Persons with hearing impairment lose this ability to hear during gaps in the presence of a masking noise. This is particularly true of elderly people. This ability to hear weak sounds during brief gaps in a more intense masker gradually decreases as hearing loss get worse. The common reason for decreased gap listening ability is that even the so called normal listeners lose some of gap listening ability as the signal to noise ratio increases. Increased signal to noise ratio is commonly needed for hearing impaired persons to understand speech.

There is another aspect of temporal resolution that is the ability to use the information contained within

the cycle-by-cylce timing of the wave form at any point on the basilar membrane. This is also known as the temporal fine structure of the wave form. Those persons who are unable to sue it are also least able to understand speech during the gaps in a masking noise. This decreased ability to use temporal fine structure may be caused by reduced precision in the timing of neural firing.

Hearing aids can help a little bit in compensating for decreased temporal resolution ability. Fast acting compression, where the gain is rapidly increased during weak sounds and rapidly decreased during intense sounds will make the weaker sounds more audible in the presence of preceding stronger sounds and could make them slightly more intelligible. This can also make unwanted weak background noises more audible.

Physiology of hearing loss

Abnormalities occurring within the outer / middle ear can cause conductive hearing loss. These abnormalities include:

1. Atresia of external auditory canal

2. Perforated ear drum

3. Ossicular chain fixity / discontinuity in the middle ear

4. Fluid accumulation inside the middle ear following infections

Inside the cochlea the inner hair cells or outer hair cells or both could cease to function normally. This can occur in some portion of the cochlea or can cover the entire extent of the cochlea.

If only the outer hair cells cease to function normally, then thresholds of hearing are elevated, dynamic range of hearing is reduced, and frequency and temporal resolution are both degraded. If only the inner hair cells cease to function then thresholds of hearing is elevated, but frequency resolution remains normal or close to normal.

The timing of signals within the brain-stem could become less precise due to a reduced number of functioning inner hair cells or a reduced number of synapses connecting to each inner hair cell. When the inner hair cells cease to function then it is common for the spiral ganglion cells to which they are connected to die progressively during the course of six months.

Sometimes the hair cells could function sub-optimally because the cochlear battery (stria vascularis) generates insufficient voltage. Hearing loss caused by inadequate stria operation is called as strial sensorineural hearing loss.

Another cause of hearing loss within the cochlea is a change that occurs to the physical properties of cochlear membranes (increased stiffness / decreased stiffness) can cause cochlear conductive hearing loss. Any defect that interferes with the conversion of vibrations in the cochlea to nerve signals is known as sensorineural hearing loss.

When the cochlea is normally functioning and there is a defect in the connection to auditory nerve or defective transmission along the auditory nerve, then the hearing loss is referred to as neural hearing loss. When the outer hair cell functions normally,

and either the inner hair cell of their connection to the auditory nerve, of the auditory nerve itself are defective. This defect is known as the auditory neuropathy spectrum disorder. This condition is common in children who are born with the condition or due their spending long duration in neo-natal intensive care unit.

If the inner hair cell in some region of the cochlea completely stops functioning and ceases transmission of information to the auditory nerve then that portion of the cochlea is known as the dead region. Tests to detect these dead regions are available.

Sensorineural hearing loss is considered to be mostly caused by defective Inner hair cells / outer hair cells function and should ideally be called as sensory hearing loss. But it is commonly termed as sensorineural hearing loss as there could be associated deficit in auditory nerve conduction because of loss of synapse in the spiral ganglion. Causes of hearing loss has been simplified under three categories:

1. Conductive hearing loss

2. Sensorineural hearing loss

3. Auditory neuropathy spectrum disorder.

Hearing deficits that occur in combination

The various aspects of hearing loss which include:

Decreased audibility

Decreased dynamic range

Decreased frequency and temporal resolution

Occurrence of dead regions

They all cause a reduction in intelligibility of sound. Any combination of these can cause a hearing impaired person to understand much less than a normal hearing impaired person to understand much less than a normal person. A hearing impaired person needs a better signal to noise ratio for speech understanding than normal person.

Signal to noise ratio deficit can occur in auditory processing disorders. These disorders include disorders of brain-stem, mid-brain, or auditory cortex and can exist independently of any peripheral hearing loss and can be a consequence of impaired cochlea sending deficit signals to the brain-stem.

One of the auditory processing disorder that has been studied extensively is the ability to separate a target from competing speech on the basis of direction of arrival. The deficit in signal to noise ratio observed in hearing impaired persons is much greater when the target speech and the competing sounds are spatially separated, and they all come from the same direction. This is known as spatial processing disorder.

The magnitude of signal to noise ratio deficit for spatialized sounds can be measured clinically with the Listening in spatialized noise sentences test. This spatial deficit is exacerbated by aging.

Binaural auditory processing system ensures that normal hearing individuals to focus their attention in one direction and suppress sounds from other directions. This feature is adversely affected by cochlear distortions that occurs in sensorineural hearing loss. It is also pointed out again to reinforce the fact that the signal to noise ratio required for a given level of speech intelligibility increases as the amount of sensorineural hearing loss increases.

Signal to noise ratio (SNR) deficit is greater:

1. If the competing signal with the talker fluctuates greatly in amplitude, as is the case with single competing speaker.

2. If the hearing impaired person is significantly older than the normal people with whom this value is compared with.

3. If the speech and competing signals come from different directions.

4. If the competing signal has spectral gaps that cause it to be a more effective masker for people with sensorineural hearing loss (causing reduced frequency selectivity) than for people with normal hearing.

If a speech in noise test is administered to a patient uses a non fluctuating noise with a smooth spectrum, and there is no spatial differentiation of signal or noise, the degree of SNR deficit measured will underestimate the deficit that the patient will experience in real life when compared to those with normal hearing.

Experiments reveal that for every 10-dB increase in four frequency average hearing loss requires a 1 dB to 3 dB increase in SNR to keep speech intelligibility in noise constant. Largest of these values are known to occur when the target and competing sounds are spatially separated. From this it should be understood that the ability to separate speech from noise using spatial clues could be one of the major cause for hearing impaired person still having trouble understanding speech despite using a good hearing aid.

When normal hearing individuals are deprived of audibility by adding noise to simulate hearing loss, their speech intelligibility scores deteriorate to levels similar to those of people with equivalent mild to moderate hearing loss. But in a person with significant sensorineural hearing loss the SNR deficit remains despite amplification by the hearing aid. Hence greater the hearing loss greater will be the SNR loss.

As far as conductive hearing loss is concerned the situation is rather simple. Conductive hearing loss cause a simple attenuation of sound, hence the provided hearing aid can amplify the sound adequately for the normal cochlea to resolve sounds entering it and hearing will return to near normal when hearing aid is used. Hearing aids are very beneficial to patients with conductive hearing loss. When conductive hearing loss increases, the proportion of sound reaching the cochlea by bone conduction also increases, consequently the inpur reaching both the ears are more or less similar and the brain finds it difficult to combine these two signals to selectively attend to the sounds coming from one direction decreases. In this scenario hearing aids increase

the proportion of sound received by air conduction, but will large conductive losses some mixing within each cochlea of the signals reaching from the left and right hearing aids are likely to remain, this reduces spatial listening abilities compared to normal hearing.

When the cochlea ceases to send signals to one frequency region due to the presence of dead regions, the nerve cells within the auditory cortex that normally receive these signals are likely to break their existing connections and instead respond to adjacent frequencies that are being more effectively delivered by the cochlea, or even to a different modality like vision. This feature is known as neural plasticity which has profound implications in auditory rehabilitation. The affected person may require many months to fully learn to make use of the amplified sound.

If a person with long term profound hearing loss receives a cochlear implant, then the ability to use the signals coming from the cochlea could remain minimal for ever and neural plasticity is therefore limited in these patients.

History of Hearing Aids

The very beginning of augmentation of sounds and presenting them to the ears began during the 17th century. Before that humans who had difficulties in hearing were condemned to the world of silence. The first device that assisted a person with diminished hearing was the ear trumpet that was introduced during the 17th century. It was so popular that many manufacturers took up manufacturing of these devices. During 18th century these devices were extremely common. To make these devices portable collapsible conical ear trumpets were manufactured for specific discerning clients.
During this period many well known models were introduced.

They include:

Townsend Trumpet – This was designed by the deaf educator John Townsend)
Reynolds Trumpet – This was specially built for the painter Joshua Reynolds
Daubeney Trumpet.

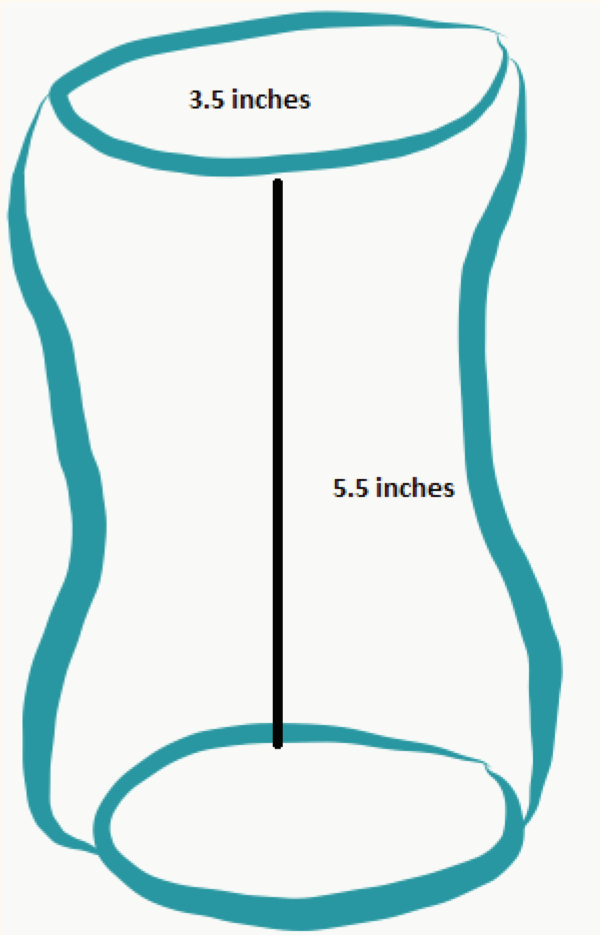


Figure 1.2 showing Townsend trumpet design along with its dimensions.

Commercial production of ear trumpet was first begun by Frederick C. Rein in London in 1800. He also managed to design hearing fans and speaking tubes. These instruments were portable and still managed to function well in amplifying sound. These instruments however had a drawback, they were heavy and had to be supported physically from below. Later small hand-held ear trumpets were designed and were used as hearing aids.

Rein's Flower vase receptacle which was designed to serve as sound collector became a curiosity. This object resembling a flower vase would be placed on a table and would be filled with flowers. It had six openings which served as sound collectors.

A tube is inserted into the base of the flower vase and the same would be held close to the ear to augment the sound levels. This innovation by Rein prompted the King of Portugal John VI (1819) to commission him to design an acoustic chair for him. He designed a throne that had beautifully carved arms that resembled the open mouths of Lions. These openings served as a receiving area for the sound waves and the same was transmitted to the back of the throne via a speaking tube into the king's ear.

Rein also pioneered in designing different types of hearing aids during his time. Notable among them was the design of "acoustic headbands" that could be artfully concealed within the hair of head gear.

Ivory ear trumpet:

This was specially made and was used by Admiral Sir John Borlase Warren, Bart in 1822. This instrument consisted of a trumpet, an earpiece and a series of cylinders forming a tube. This trumpet was made using ivory.

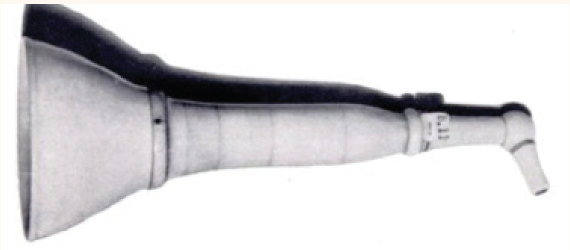


Image 1.3 showing Ivory ear trumpet

Electronic hearing aids:

The first electronic hearing aids were constructed after the invention of telephone and microphone in the 1870's and 1880's. The technology used in telephony demonstrated that acoustic signals can be altered and conducted electrically. The very first electronic hearing aid known as the Akouphone was designed by Miller Reese Hutchinson in 1898. In this instrument a carbon transmitter was used to make the device portable.

The carbon transmitter amplified the sound by taking a weak signal and used electric current to make it a strong signal. These electronic hearing aids reduced greatly the size of the device.

Siemens of Germany was the first company to manufacture these electronic hearing aids. The first hearing aid manufactured was rather bulky i.e. about the size of a large cigar box and had a speaker that would fit in the ear.

Vacuum tube hearing aids:

This technology patented by Naval Engineer Earl Hanson in 1920 used a telephone transmitter to turn speech into electrical signals. After the signals were converted into electrical waves, it would be amplified when it moved to the receiver. This hearing aid weighed roughly 7 pounds and was light enough to be carried. Marconi in England and Western Electric in the US began marketing these hearing aids in 1923.

The first wearable hearing aid using vacuum tube technology went on sale in England in 1936. Then came the patenting of automatic gain control technology by Multi-tone.

Many technological advances took place during the second world war and the same was used in designing hearing aids. The technology involved was in miniaturization. Zenith produced a pocket-sized miniature hearing aid soon after completion of the second world war.

Age of transistors:
Development of transistors in 1948 by Bell laboratories pioneered improvements to the hearing aid technology. These transistors developed by William Shockley replaced vacuum tubes since they tended to get hot and was also fragile. These transistors served to reduce the size of the hearing aid still further. They also played an important role in the development of miniature carbon microphones.

Since the process of integrating transistors to hearing aid took place rather quickly an important aspect was overlooked. These hearing aids ceased to function after a few weeks of usage, this happened because the transistors failed to function when they get damp. Further innovation occurred to tackle this problem. They started coating the transistors which prevented their malfunction if they happened to get damp. The use of integrated circuit board still made the size of hearing aid exceedingly small.

Time line of hearing aids

13th - 19 th century:

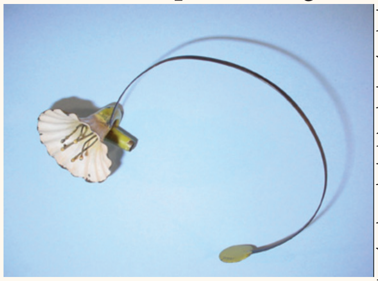
Era of animal horns and trumpets. Animal horns of varying sizes and shapes were initially used as hearing augmentation devices. Then trumpets were designed using commonly available materials those days. Ear trumpet that was designed using ivory was famous during this period.

13th - 19th centurty (animal horns to trumpets)

As early as the 13th century persons with hearing loss especially the aristocrats and kings started using hollowed out horns of animals like that of cows and rams as primitive hearing aid devices. Only during the 18th century that the modern hearing aid design (ear trumpet) became popular. These devices did not amplify sound but served as collectors as they collected sound and funneled it to the ear.

Image 1.4 showing trumpet head band hearing aid designed by Rein who should be considered as a pioneer in hearing aid design.

The trumpet headband was designed by F.C. Rein of England in 1850. The metal headband could be worn either on top of the head or worn around the base of the neck.
Even those days hearing aid designers used innovative techniques to disguise these devices. Shown



here is one such design where the sound collector is disguised as a flower. This device was known as the Floral Aurolese phone. This was again designed by Rein in 1802 who was forefront in hearing aid design those days.

19th - 20th century: The era of electronic hearing aids

The invention of telephone combined with the application of electricity in 19th century had a huge impact on the development of hearing aids and other hearing assistive devices. Persons with hearing loss suddenly realized that they were able to hear conversation better through the telephone receiver when held close to the ear than when being seated in front of the speaker. Thomas Edison who had hard of hearing introduced a carbon transmitter for the telephone that amplified electrical signal and increased the decibel level by about 15 dB. This particular invention paved the way to the use of carbon transmitter for hearing aids.

Acoustic chairs

This is a classic and innovative example of incorporating a hearing device within an every day object (integrating hearing accumulating device into a chair). These chairs were commonly used by kings who had difficulty in hearing. Nature treats every human being as equals and it is no surprise that kings also had hard of hearing. They of course had the wherewithal

to create assisting devices. These acoustic chairs conveyed sound discretely through the armrests to a hearing tube placed at the back of the chair. King Goa chair had the most ingenious design and was created by the master designer F.C. Rein for King

John VI of Portugal. He extensively used this chair while ruling from Brazil. This chair was equipped with a large receiving apparatus concealed beneath the seat. Its hollow arms were elaborately carved to represent the open mouth of lions which acted as receivers of sound.

Curtis acoustic char was equipped with a large trumpet alongside the chair to transmit sound to the user's ear. Curtis also remarked that this chair had a great advantage in that the person sitting in the chair was not subjected to the "unpleasant bad breath emanating from the speaker" since the speaker was required to address the person in the chair from the opposite side.

Curtis' Acoustic Chair.

1921-1952: Era of Vacuum tube technology:

Starting from 1920's hearing aids using vacuum tube technology were able to increase the sound level by as much as 70 dB. These levels were possible because vacuum tubes controlled the flow of electricity better than carbon. The only problem of these devices happened to be the size. Earlier hearing aids using vacuum tube technology were very large in fact about the size of a filing cabinet and hence were not portable.

By the year 1924 the size of the vacuum tube hearing aids got reduced so all the components could fit in a small wooden box with the receiver held close to the ear. Improvements in technology continued and in 1938, when Aurex introduced the first truly wearable hearing aids which consisted of an earpiece, wire and receiver that could be clipped to the user's clothing. This model required the use of a battery pack that needs to be strapped to the user's leg. World war II in the late 1940's saw the introduction of circuit boards and button sized batteries. This technology allowed batteries, amplifier and microphone to be combined into one portable, pocket sized unit.

Mid-20th century: Transistor technology revolution

The aim of designing a portable hearing aid that can fit into a pocket got a start in 1948 when Bell Telephone labs invented the transistor. This is a switch that controls the movement of electrons and thus electricity. These devices can start and stop the flow of current and also control the current voltage there by making it possible to have multiple settings in one device. Norman Krim in 1952 created the first transistor for hearing aid companies thus enabling the hearing aids to be made smaller and could finally be worn behind the ear.

This new technology was capitalized by Otarion electronics which managed to embed hearing aid into eyeglasses.

Late 20th century: Transition from analog to digital hearing aids

Hearing aid manufacturers developed the ability to make transistors from silicon thus enabling hearing aids to become more powerful and more smaller. This technology was introduced first by Zenith Radio in 1960 and in these versions the microphone went in the ear and was connected by a small wire to an amplifier and battery unit that was clipped to the ear. This technology was popular till 1980 till digital processing chips were introduced into the hearing aid. Zenith Radio was the first to use this technology and created hybrid digital-analog models. In 1996 hearing aids fully became digital.

21st century the era of high tech hearing aids

By the year 2000, hearing aids had the ability to be programmed allowing the user to customize the aid to their requirement. It offered lots of flexibility including fine tuning. Currently only digital hearing aids are predominant in the market.

Acoustic Measurements

Basic physical measures to measure sound include:

1. Frequency

2. Period

3. Wave length

4. Diffraction

5. Pressure

6. Sound pressure level

The terms waveform and spectrum should be understood before proceeding further.

Frequency:

This term is defined as how many times per second a sound wave alternates from positive pressure to negative pressure and back to the starting value. Frequency is usually measured in cycles per second / as Hertz (Hz) or kilo Hertz (kHz).

Period:

This is the time taken for a repetitive sound wave to complete one cycle. Period is measured in seconds / milliseconds and is equal to one divided by the frequency. (It is the time that takes for the same part of the wave's regular repetition to pass by our measurement point again).

Phase:

This value refers to the timing of the sound, or one component of a sound, relative to some other aspect of the sound relative to another sound. One complete period corresponds to a phase shift of 360 degrees. Two sounds are out of phase when their wave forms are proportional to each other but have opposite polarity. This corresponds to a phase shift of 180 degrees at all frequencies.

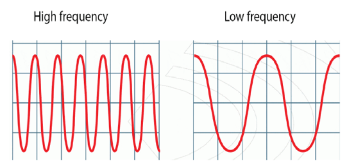


Image 1.5 showing high and low frequency audio curves.

Wavelength:

This term refers to the distance a sound wave travels during one period of the wave. It is measured in meters and is equal to the speed of sound(345 m/s) divided by the frequency of the sound. Low frequency sounds have large wavelengths (several meters) and high frequency sounds have small wavelengths (a few centimeters).

Diffraction:

This describes the way in which a sound wave is altered by an obstacle. When a sound meets an obstacle, like a head, the size of the wavelength compared to the size of the obstacle determines the end result. Sounds with wavelengths smaller than that of the obstacle cannot bend around the obstacle and hence could cause a sound shadow to occur on the side of the obstacle away from the source. In other words the sound is attenuated. These obstacles can also cause the sound pressure to increase on the side closest to the source.

Sounds that have wavelengths much larger than an obstacle will flow smoothly, without attenuation, around the obstacle, giving much the same sound pressure at all points around the obstacle.

One can hear a conversation that is happening in the next room even though the source cannot be seen. This is attributed to the phenomenon of diffraction, where sound waves bend and spread as they go through the doorway between the two rooms. Diffraction only occurs when the wavelength is close to the size of the opening or object.

Pressure:

This is described as force per unit area sound wave exerts on objects that gets in its way (i.e. ear drum). It is measured in Pascals (Pa), mPa.

Sound pressure level (SPL):

This is defined as the number of decibels by which any sound pressure exceeds the arbitrary, but universally agreed reference sound pressure of 20 micro pascals. When pressure doubles, the SPL increases by 6 dB; when pressure increases ten times, the SPL increases by 20 dB.

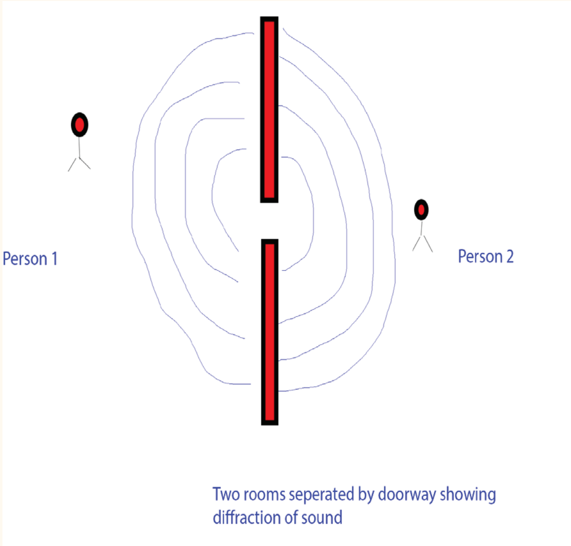


Image 1.6 showing the concept of diffraction of sound.

RMS:

This stands for the root-mean-square value of a signal. It is actually a way of representing, with a single number, the strength of a fluctuating signal over a certain time. This value reflects the average power of a signal.

Waveform:

This value describes how the pressure of a sound wave varies from moment to moment of time. The waveform of a pure tone for example is a sinusoid.

Spectrum:

This describes the mixture of pure tones that when added together, produce a particular complex sound over a specific period of time. A complete spectrum specifies the amplitude and phase of every pure tone component in the complex sound but as far as audiologist is concerned it is only the amplitude spectrum that is important.

When the complex tone is periodic (i.e. each cycle looks like the preceding cycle), the pure tone components are called harmonics. The frequencies of these harmonics occur at integer multiples of fundamental frequency, which is the frequency at which the complex wave itself repeats. A Fourier analysis is the mathematical operation that enables the spectrum to be calculated if the waveform is known. A spectrum and a waveform are two different ways of describing the same sound.

Octave band & one-third octave bands:

These are frequency regions one octave and one-third octave wide respectively. The spectrum of acoustic signals is often analyzed by filtering the signals into adjacent octave or one-third octave bands. An octave corresponds to a doubling of frequency.

Critical bands:

These are frequency regions within which it is difficult for the ear to separate sounds of different frequencies. Sounds separated by more than a critical band are likely to be separately recognized by the brain by persons with normal hearing. The cochlea processes sounds through auditory fibers centered at each and every place in the cochlea. These areas act as band pass filters which progressively attenuate sounds as their frequency departs from each fiber's center frequency. The width of each of these filters can be characterized by its equivalent rectangular bandwidth (ERB). For center frequencies above 1000 Hz an ERB is approximately 1/6 of an octave wide. The bandwidth in Hz therefore increases proportional to the center frequency of the band. As the center frequency decreases below 1000 Hz, the bandwidth in Hz decreases, but the relative bandwidth in octaves increases. By 100 Hz, the ERB is about 30 Hz or half an octave wide.

Impedance:

This value describes how easily sound gets conducted through a medium or how easily a medium vibrates when sound pressure is applied to it. In free air impedance is equal to the ratio of sound pressure to particle velocity (the velocity at which particles in the medium vibrate back and forth as the sound wave propagates). The impedance of a medium has a constant value that depends only on the physical characteristics (density and elasticity) of that medium.

In tubes impedance is defined rather differently which is equal to the ratio of sound pressure to volume velocity. Volume velocity is defined as particle velocity multiplied by the cross sectional area of the tube. This can also be described as the total quantity of sound flowing back and forth through any plane perpendicular to the length of the tube.

Linear amplification and gain:

The gain of any device relates to the amplitude of the signal coming out of the device to the amplitude of the signal going into the device. Gain is calculated as the output amplitude divided by the input amplitude. If these signals are electrical then they are measured in Volts and acoustic signals

are measured in Pascals. If an input signal of 20 mPa is amplified to become an output signal of 200 mPa then the gain of the hearing aid would be ten times. This is actually known as linear amplification. This makes every input signal bigger by multiplying the input signal by a fixed amount.

For the sake of convenience, the input and output signal amplitudes are expressed as a level in decibels (dB SPL). Gain is calculated as the output level minus the input level and is expressed in decibels. For example if the input signal of 40 dB SPL becomes amplified to 80 dB SPL then the gain is calculated as 40 dB.

The relationship between input and output SPL for a particular frequency can be studied using an input-output diagram (I-O diagram). Amplification is considered to be linear if the signal intensity of different frequencies are amplified by the same value. The behavior of a linear amplifier is not affected by how many signals it is amplifying at the same time.

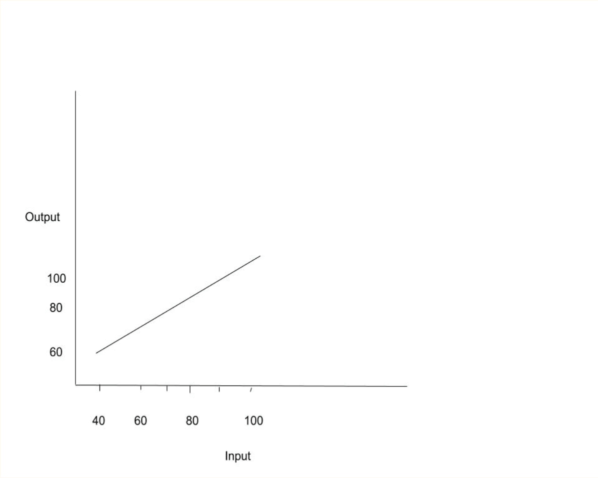


Image 1.7 showing IO chart

Measurements of gain-frequency response will be meaningful if the signal used is specified.

Saturation sound pressure level:

All amplifiers become nonlinear when the input or output signals exceed a certain level. This occurs because amplifiers are unable to handle signals larger than the voltage of the battery that powers the amplifier. It is also desirable to limit the maximum output of the hearing aid to be even less than the limit imposed by the battery voltage and below the limit imposed by the receiver. The highest value of SPL that a hearing aid can produce is known as SSPL (saturation sound pressure level). This value varies with the frequency and is a useful measure.

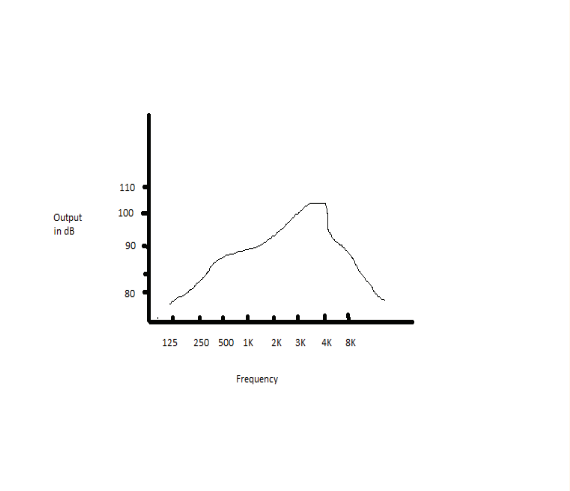


Image 1.8 showing saturated sound pressure level frequency curve

Hearing aids are meant to be used in ears, hence the best place to measure the output of a hearing aid is in the ear canal of the aid user. This is not possible sot the only practical way is to use it in couplers and measure the value. Couplers are soft,

thin probe tune attached to a microphone. It is also desirable to be able to measure a hearing aid in a way that does not require it to be mounted in a person's ear. Coupler becomes useful for this purpose.

Couplers are small cavities. The hearing aid connects to one end of the cavity, and the other end of the cavity contains a microphone which in turn is connected to a sound level meter. The commonly used coupler has a volume of 2 cc. The coupler hence goes by the name 2-cc coupler. Currently ear simulators are being used in place of couplers.

Couplers and ear simulators are indispensable for confirming that hearing aids are operating correctly. Due to individual differences in the ear canal volume and geometry and the unique way in which the hearing aid is coupled to the ear, the hearing aid's performance should be measured in the ear of each individual hearing impaired person. This process is known as the hearing aid trial.

Components of a Hearing Aid

Components of hearing aid include:

Microphone:

This component picks up different sounds in the environment and converts them to electric signal which can be understood by the signal processor. Currently available microphones can differentiate sounds such as speech and background noise and process them differently to enable seamless hearing experience. Since it converts sound waves into electrical waves it is also known as the transducer. There are two types of microphones currently available:
Directional microphones – These microphones usually pick up sounds from the front of the user. This can be useful to understand a conversation in a noisy environment.

Omni-directional microphones – These microphones pick up sounds from all directions, helping the user to get a better sense of where sounds are coming from. The currently available hearing aids use both these types of microphones in order to provide the user with the best possible listening experience.

For a perfect microphone the wave form of the electrical signal coming out of the microphone is identical to the waveform of the acoustic signal going into the microphone. Majority of microphones act in a linear fashion, so that every time the pressure of the input signal double, the output voltage also doubles, until the output reaches the highest voltage that the microphone can deliver. The ratio of the size of the output voltage to the size of the input voltage is known as the sensitivity of the microphone.

Microphones used in hearing aids have a sensitivity of about 16 mV per pascal, which means that sounds of 70 dB SPL produce a voltage of around 1mV.

Processor / Amplifier:

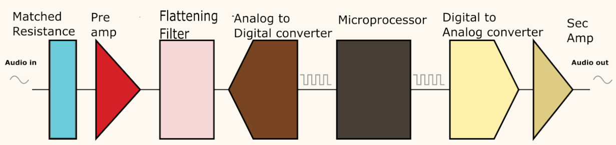
This should be considered as the motherboard of a computer. This component of the hearing aid takes the electric signals from the microphone and converts them into digital signals making it easy for manipulation. In this step the digital signal can be adjusted to the requirements of the patient, which includes the amount of amplification needed to accommodate the hearing loss. Feedback noise should be reduced or canceled. Tinnitus masking feature if any needed is added at this step. After the digital signal is processed adequately by application of various filters, the processor then converts this signal back to an analog signal. This converted analog signal is sent to the third and final hearing aid component that is the receiver.

Receiver:

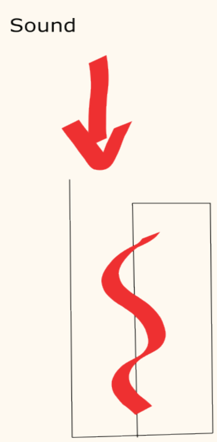
This is the final component in the hearing aid. It creates an enhanced sound wave that will somewhat meet the needs of the user. The receiver converts the signal sent from the processor to audible sounds and outputs it to the ears of the hearing aid wearer.

This portion of the hearing aid directs sound towards the patient's ear. Some of the hearing aids have the receiver placed in the ear canal, as seen in completely -in – the – canal type. Other devices have the receiver connect to a small tube that is inserted into the external auditory canal as is seen in behind the ear model hearing aid.

Image 1.9 showing various components of hearing aid



Non directional (unidirectional) Microphone:



As shown in the figure above the body of this microphone can be imagined as a box that is divided into two chambers by a flexible diaphragm in the middle. Sound enters through the small opening at the top of the diaphragm and travels to the chamber on one side of the diaphragm. This open chamber is known as the "Front Volume". The acoustic pressure exerted by the sound in this chamber causes the diaphragm to vibrate a little. The chamber on the other side of the diaphragm also known as the "Back Volume" is closed to the outside environment. An electrically charged plate is placed in the back volume very near the diaphragm (not shown in the drawing). The movement of the diaphragm induces an electrical voltage on the diaphragm which is amplified as the output signal of the microphone.

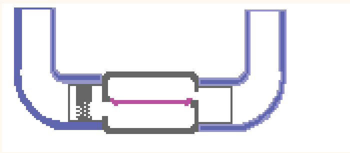


Image 1.10 showing bidirectional microphone.

Directional microphone:

This is basically the same as non-directional microphone except for the fact that an opening is made into the back volume also for another sound entry port.

This unit has two sound ports that sample the sound pressure at locations separated

by a small distance. Sound pressure travels to both sides of diaphragm. The motion of the diaphragm is determined by the difference in sound pressure on both sides of the diaphragm. If there is no pressure difference on either side of the diaphragm, then there is no movement of the diaphragm leading to zero output. However,
as the sound wave propagates from right to left there will be a time delay in the acoustic signal at the two ports. This delay causes a small difference in pressure which enables the diaphragm to vibrate causing generation of electrical voltage.
In almost all these types of microphones, there is a small resistive screen placed in the rear sound entry tube. Insertion of this screen provides an additional time delay in the acoustic signal moving to the back volume. This allows the directional response to be adjusted to suit the specific situation.

Combining tow non directional microphones to facilitate sound localization:

Two non-directional microphones are included in the hearing aid. The microphone outputs are combined in the hearing aid circuitry to form a directional pattern. The small differences in performance between these two microphones help in sound localization.

Directional microphones have more noise than non-directional microphones. It should also be stressed that directional patterns are better for hearing in noise than non-directional microphones. Directional patterns are noisier than non-directional patterns. Currently available hearing aids have the ability to switch to a non-directive pattern when needed and this switching takes place reliably. This switching can be performed manually by the user or can be set to automatic.

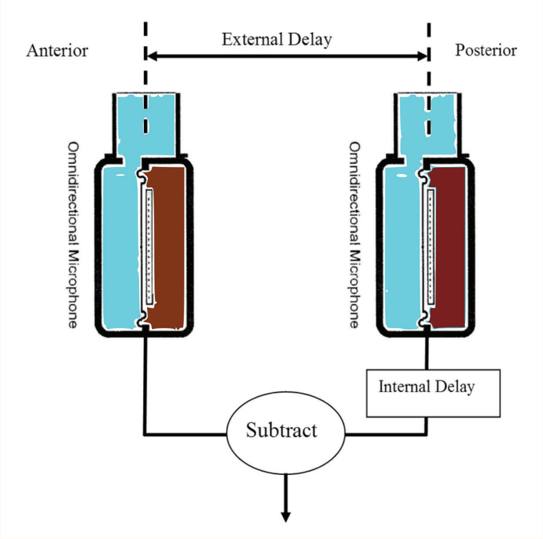


Image 1.11 showing the use of two unidirectional microphones

Other components of the hearing aid:

Ear hook / connecting tube:

This is a clear tube that connects the receiver and loops over the top of the ear as seen in the behind the ear model. In the body worn hearing aid this tube connects the

receiver with the head phone.

Battery:

This provides the power source for hearing aid components.

Vent:

This allows for airflow and prevents the feeling of a plugged ear.

Volume control:

Allows the user to make adjustments to the intensity of the sound.

Wax guard:

This is a small replaceable filter that prevents ear wax from getting into the internal hearing aid components.

Microphone mechanics:

Microphones make use of several fundamentally different types of technologies, but since the 1980's hearing aids are using only one type of microphone and it is the electret microphone. Sound waves enter through the inlet port and reach one side of a very thin and flexible plate with a metalized surface called the diaphragm. Pressure fluctuations within the sound wave cause the diaphragm to move up and down by an extremely small amount, it is invisible to the naked eye. A small air space separates the diaphragm from a rigid metal plate known as the back plate. Coated onto the back plate is a thin Teflon material called an electret. The diaphragm is held away from the back plate by some bumps in the back plate. The back plate has holes in it to allow movement of air through it.

The electret material has a permanent electric charge comprising an excess of electrons on one side of it, and a shortage of electrons on the other. When sound pressure forces the diaphragm towards or away from the electret, the changing distance between the diaphragm and the electret changes the electrical force between the opposing charges thereby varying the voltage between the back plate and the diaphragm. This mechanism turns the sound wave into electrical waves. An amplifier is built into the same container as the rest of the microphone. It amplifies this generated current and delivers it to the main amplifier built into the hearing aid. This amplifier present within the microphone is referred to as a FET, as it is made of a special type of transistor known as the Field effect transistor. It is also referred to as a buffer amplifier as it prevents the main amplifier built into the hearing aid from acting because of the loading down effect.

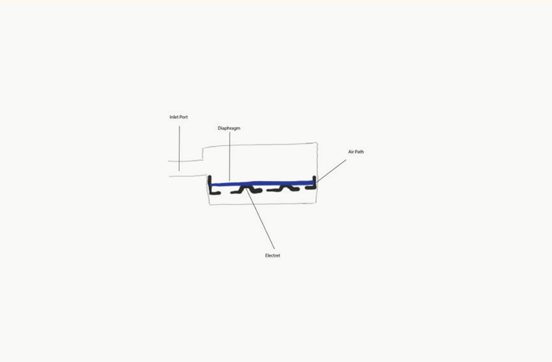


Image 1.12 showing structure of electret microphone.

Silicone Microphone:

Fully electronic microphones are also available. These are also known as the silicon microphone (solid state microphones, micro-electro-mechanical systems microphone). These microphones are made by etching away parts of a block of silicon, and depositing layers of other materials onto it. These techniques are similar to that of manufacture of integrated circuit. These microphones have not been fully perfected yet. If introduced this will reduce the cost of the hearing aid to a great extent.

Frequency response of microphones:

Electret microphones have frequency responses that are flat, variations from a flat response could occur either by design or by accident. A low cut is commonly intentionally introduced into electret microphones used in hearing aids. A low cut makes the hearing aid less sensitive to the intense low frequency sounds that commonly surround us. These low frequency sounds may not be perceived even by normal hearing people but could cause the microphone or the hearing aid to overload unless the microphone inherently attenuates them.

To achieve low cut in a microphone is rather simple. A small passage way is provided between the front and back of the diaphragm which allows low frequency sounds to impact almost simultaneously on both sides of the diaphragm thereby reducing the mobility of diaphragm to these sounds. The larger the opening greater will be the attenuation and greater will be the frequency range at which attenuation occurs. This opening also serves to equalize the static air pressure between the front and back of the diaphragm. Microphones with different degrees of low cut are used in custom hearing aids in order to achieve a good gain-frequency response for the hearing aid as a whole.

Another variation that could occur from a flat response is the result of an acoustic resonance within the microphone casing. A resonance is known to occur between the air in the inlet port and the volume of air

next to the front of the diaphragm. The mechanical compliance of the diaphragm itself and of the air behind the diaphragm could also contribute to the resonance. This resonance is also known as the Helmholtz resonance. This resonance causes a peak in the gain frequency at about 5 dB high and centered in the range between 4 - 10 kHz. The shorter and wider the inlet port, the higher is the resonant frequency, and consequently the greater is the high frequency bandwidth.

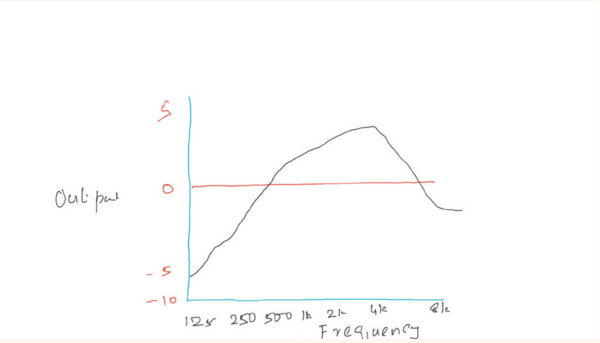


Image 1.13 Frequency response of a typical electret microphone with tubing on its input port.

Imperfections involving the microphone:

The major imperfection which involves a microphone is that they could break down if they are exposed to adverse chemical agents like perspiration. All electronic components are known to produce small amounts of random electrical noise. Microphones are no exception to this fact. Noise is caused due to the result of random motion of air molecules against the diaphragm, and partly as a result of random electrical activity within the internal microphone amplifier. The noise when it gets amplified by the main hearing aid amplifier it becomes audible to the user.

Another imperfection of microphones is that they are sensitive to sounds as well as vibrations. If the microphone is shaken the inertia of the diaphragm causes is to move less than the outer case of the microphone. Consequently, the diaphragm and the case move relative to each other generating a voltage reflecting the magnitude and frequency of the vibration. The immediate sequel to the sensitivity to vibration is that they will be amplified into an annoying sound. Even rubbing of hearing aid casing with the clothing is enough to generate noise. Direct vibrations from the body as it occurs when running on a hard surface may also be audible as a thumping noise.

Another issue is that when the hearing aid receiver operates it creates vibrations as well as sound. The microphone picks up some of these vibrations and converts them into electric signal which are amplified by the hearing aid and passed on to the receiver. If the vibrations from the receiver is strong enough and the gain of the hearing aid is high then this could cause a feedback loop leading on to audible oscillation which is of low fre-

quency. Designers of hearing aid avoid this problem by careful mounting and placement of the microphone and receiver.

Displacement of transducers from their position commonly occurs in (ITE model hearing aids) In the canal aids or completely in the canal aids because of their small size and custom construction. This can be repaired by repositioning the transducers.

Another imperfection can occur due to poor design of the hearing aid. If the microphone is mounted with a long thin tube on its inlet port then it could cause Helmholtz resonance causing a larger peak in the gain frequency response and a rapid decrease in gain for frequencies above this peak frequency.

Another issue that can arise is that microphones could be subjected to wind noise. When wind hits an obstacle like the head / pinna it could cause turbulence which can be picked by the microphone of the hearing aid causing a audible hissing noise. Even moderate wind speeds can produce very high SPL's at the microphone input thereby overloading the microphone.

Keeping the microphone inlet away from the wind flow could minimize the amount of wind noise. One effective but cosmetically unacceptable method is to place some plastic foam over the microphone port. Better way would be to place the microphone port inside the ear canal as is seen in complete inside the canal hearing aids. Another option is to have a large microphone port covered by a mesh screen or dome. This reduces noise as the pressure fluctuations across the large opening partially cancel each other. This does not happen if the opening is small. Another effective way is for the patient to wear a light scarf with sufficient opening for hearing. This will prevent wind from directly hitting the hearing aid and the pinna.

Directional microphones:

These microphones suppress noise coming from some directions, while retaining good sensitivity to sounds arriving from one direction. Directional sensitivity of microphones is indicated on a polar sensitivity pattern. This pattern of response is also known as the cardioid because of its heart shape. In real life situations unwanted noise arrives more or less equally from all directions. Even if the noise originates from only one / two sources, room reflections cause the energy to arrive at the aid wearer from all directions. If the wearer of the hearing aid is standing close to the person with whom they are conversing, the wanted signal will arrive mostly from directly ahead. A good directional microphone should have maximum sensitivity for sounds arriving from directly ahead, but the sensitivity averaged across all other directions is referred to as the directivity index (DI).

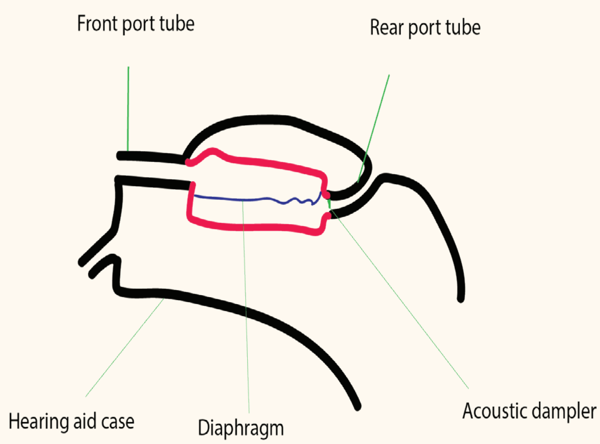


Image 1.14 showing the components of a directional microphone

Directionality occurs because the head and pinna attenuate the sound when they come between the source and the microphone, and boost the sound when the microphone is positioned between them and the source. The boosting and attenuating effects of head diffraction increase in magnitude as frequency rises.

Directional microphones are most commonly used in Behind the ear hearing aids. When hearing aid wearers need to hear people behind them, it is counterproductive to sue a directional microphone. In order to solve this problem the hearing aid should include both a directional microphone and an omni-directional microphone and the user should select which one is suitable as per the situation. Another approach is to incorporate two separate omni-directional microphones, each with one inlet port. When an omni-directional sensitivity pattern is needed, the output of one microphone is selected or the two outputs are added together. When a directional sensitivity pattern is needed, the two microphones are used in a different combination, this is also known as dual-microphone technique. In this technique, the output from the second microphone is electronically delayed and then subtracted from the first microphone, making an exact electronic equivalent of the processing that happens acoustically in a single directional microphone.

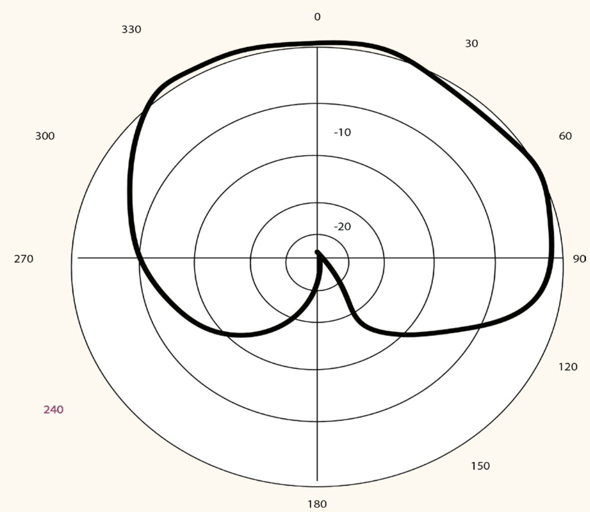


Image 1.15 showing Directional sensitivity (in dB) of a microphone with a cardioid sensitivity pattern.

Microphones can be designed have a sensitivity that depends on direction of arrival of sounds to both sides of the diaphragm from two separate inlet ports. The directional properties of the microphone depends on two delays:

1. The maximum external time delay is the time taken for sounds arriving from the front or rear to get from one inlet port to the other and is approximately equal to the distance between the ports divided by the speed of sound in the vicinity of the head.

2. The internal time delay arises because the rear port contains an acoustic damper or resister. This combines with the cavity at the back of the diaphragm to create a low pass filter that passes most of

the amplified frequencies without attenuation with some delay. This delay is inherent to all filters.

Sound coming from the rear direction gets to the rear port earlier than the front port. Sound entering the rear port is delayed as it passes through the internal low pass filter. If the internal and external delays are equal, then the sound from the rear will reach both sides of the diaphragm at the same time and there will be no net force on the diaphragm. In this state the microphone is insensitive to sounds coming from other directions.

Amplifiers:

The function of amplifier is very simple. It converts a small electrical signal into a larger electrical signal. Since the microphone has already converted the sound into electrical voltages and currents, the amplifiers can do the following things:

1. Amplifiers can make the voltage larger, without affecting the current.

2. Amplifiers can make the current larger, without affecting the voltage.

3. Amplifiers can make both the voltage and current larger.

All these options cause the signal to have more power as it comes out of the amplifier. Amplifier takes the power from the battery and transfers it to the amplifier output in a manner controlled by the input signal. The output waveform is simply a larger version of the input waveform.

Technology used in the amplifier:

Main element of an analog amplifier is the transistor. Even though a single transistor is capable of providing amplification, there are many transistors inside the amplifier. Transistors can be connected to the resistors together helps the amplifier to amplify sounds in a proper and efficient manner. These multiple transistors and resistors are embedded using photographic and chemical techniques into an integrated circuit.

Transistors are made using either bipolar and CMOS (complementary metal oxide semiconductor). Each of these types have their own advantages. Bipolar transistors tend to have lower internal noise whereas the CMOS transistors tend to use less battery power. Both these types are used in hearing aids and both could have acceptably low noise and power consumption.

Within the hearing aid to fulfill its applications there could be a few hundred to few thousand resistors depending on the complexity of the hearing aid. In most hearing aids, the IC's are mounted onto the circuit boards with electrical connections already printed on them.

Peak clipping and distortion

Pre-requisites of an ideal amplifier include:

1. It should have the gain-frequency response required

2. It should not generate internal noise

3. Should not distort the signal no matter how large the input signal happens to be.

Amplifiers inside hearing aid more or less fulfill these criteria. Deviation from the ideal occurs when the signal is too large for an amplifier to handle properly. Amplifiers cannot produce signals larger in voltage than the specified maximum. This maximum is usually equal to, or related to, the battery

voltage. If the biggest signal in the amplifier is near the maximum the amplifier will clip (remove) the peaks of the signal.

Peak clipping is of two types:

Soft clipping - This is less aggressive in that it reduces the peaks gradually rather than cutting them off by gently transitioning between the unclipped section of the wave form and the clipped section. Soft clipping begins to reduce the peaks before the threshold level is reached and then progressively increases its effect as the input level increases so that the threshold is never exceeded. This is less harsh than that of hard clipping and is known for adding warmer harmonic distortion. It also provides a smoother and more musical sounding distortion that retains more punch. This is an ideal effect in an hearing aid.

Hard clipping - This introduces a more aggressive distortion effect. It functions like a limiter by chopping off the peaks at the set threshold rather than smoothly reducing them. Excessive use of hard clipping creates a harsh distortion that is often unpleasant to hear.

The output after peak clipping that too after hard peak clipping will not be a sine wave, and it contains components at frequencies not in the input signal. These additional components are known as the distortion products. When the input is a sine wave, the distortion products occur at frequencies that are harmonics (integer multiples) of the input frequency. This distortion is known as harmonic distortion. If the peak clipping is symmetrical, the distortion products occur only at odd harmonics of input frequency. Distortion degrades the quality of speech and other signals if present in excessive amounts.

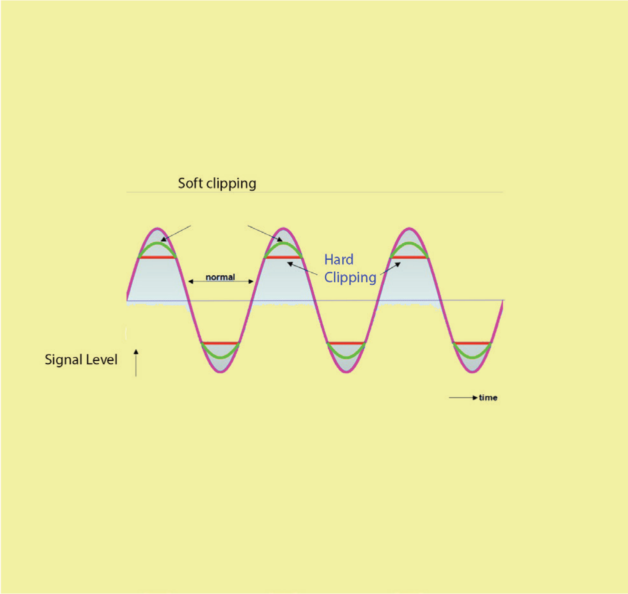


Image 1.16 showing peak clipping curves

Types of amplifiers:

All amplifiers are not the same and there is a clear distinction in the way in which their output stages are configured and operate. The main operating characteristics of an ideal amplifier are linearity, signal gain, efficiency and put output. In real world scenario there is always a trade off between these parameters.

One of the methods used to distinguish the electrical characteristics of different types of amplifiers is the classification into different classes. The term Amplifier Class is used to differentiate between the different types of amplifiers.

Amplifier classes are mainly lumped into two basic groups. The first group is classically controlled angle amplifier and are categorized as A, B, AB and C. Class A amplifiers are the most common type used. They have excellent linearity, high gain and low signal distortion levels if designed correctly. Class A

amplifiers have an efficiency of about 30% (they can amplify only by a factor of 30%) making it impractical for using in high power amplifications. This type of amplifier uses 100% of the input signal. In a class B amplifiers the device conducts for 180 degrees of the cycle. This will be a problem if only one device is used, hence two devices should be used. This class of amplifiers have an efficiency of about 60%. Class C amplifiers use less than 50% of input signal and the conduction angle is less than 180 degrees. Distortion is also high. The efficiency of this class of amplifiers is about 70%. Class D amplifiers can operate from a digital signal source without requiring a digital to analog converter.

Compression amplifiers:

Persons with sensorineural hearing loss have their dynamic ranges reduced. It is much less than that of persons with normal hearing. These patients require less amplification for intense input sounds than for weak input sounds. This condition is satisfied by compression amplifiers. This amplifier turns down its own gain as the input value increases. The compression amplifier should leave the fine detail in the wave form unchanged. The compression amplifier is also known as the automatic gain control / automatic volume control.

The compressor functions as if a human finger is introduced inside the hearing aid which rapidly and smoothly turns the volume control as soon as the output signal has reached a critical level.

Currently available amplifiers are digital amplifiers and sound here is represented as an ever changing strings of numbers. Another component of the hearing aid known as the Analog to digital converter (ADC) which changes the analog voltage coming out of the microphone into these strings of numbers which the digital amplifier uses to amplify. Digital technology has a greater predictability of operation, there is less internal noise and has the ability to perform complex operations with small integrated circuits which consume less power. As in the analog hearing aids, digital hearing aids use microphone to convert sound to an analog voltage. Another device known as the analog to digital converter changes this voltage to a series of numbers. The digital processor present inside the hearing aid performs complex arithmetic on these numbers to manipulate the sound.

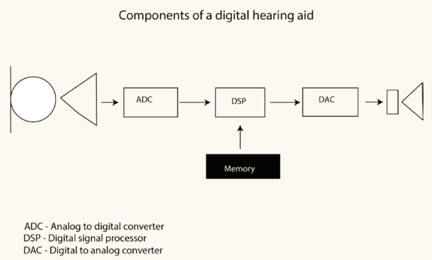
Analog amplifiers an electric voltage analogous to the acoustic sound pressure is created. When the sound pressure increases from one moment to the next then the electric current produced also increases. This technology was first introduced in the telephony and was used in earlier hearing aids. Digital technology has made all these obsolete.

Analog to digital converters:

This component is present in all digital hearing aids. Its job is to change the analog electrical voltage coming from the microphone into numbers so that the signal can be digitalized. Sampling is the first step in this process. In the sampling process a signal's size is first noted at regular intervals ignoring the value of the signal. The signal sampling should be preformed rather often in order to ensure that the sampled signal is a good representation of the original signal. It has been proved that no information of the original signal is lost provided the sampling frequency (sampling rate) is greater than twice the highest frequency component present in the complex signal. If the selected hearing aid should faithfully amplify signals up to 10 kHz, the sampling frequency has to be at-least 20 kHz, meaning the waveform should be sampled every 1/20,000 of a second or every 5o milliseconds.

The hearing aid needs to have a low pass filter to make sure that the signals going into the

Image 1.17 showing the components of a digital hearing aid.



Analog-to-digital converter are indeed lower in frequency than half the sampling frequency. This filter is known as the anti-aliasing filter. This filter ensures that if a signal component with a frequency greater than half the sampling frequency gets into the analog-to-digital converter, the hearing aid will amplify this signal as though it has a frequency lower than half the sampling frequency. Signals with excessive frequency alias themselves down to lower frequencies.

The digitalized code values are broken up into bits. Bit is a contraction of the words binary digit. Binary digits are allowed to have only two values (0 or 1) and bigger numbers are made by combining them using multipliers of 2,4,8 and 16 and so on.

Digital signal processors

This unit inside the digital hearing aid performs the following functions:

1. Protects the ear against noise

2. Modification of sound is possible since the signal is digitalized using mathematical algorithm.

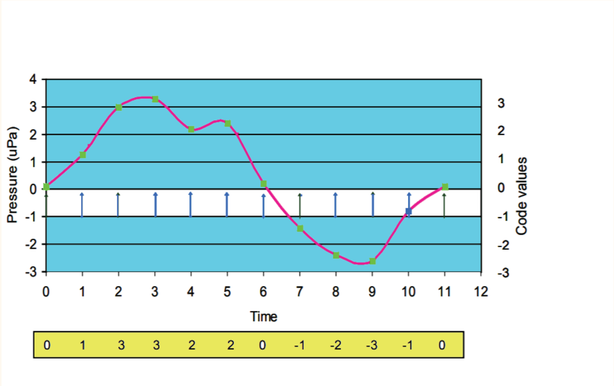


Image 1.18 showing Analog pressure waveform plotted as curve. Sampled values are shown as dots. The box below shows the digitalized value of the signal.

Digital signal processors are of two types:

Hard wired digital processor:

In this type different parts of the processor perform some specific function like a compressor or a filter. These blocks are connected together in a particular fixed order. Sound passes through various blocks of the processing in a particular order and each block will perform only one function either filtering or compression that it has been designed to perform. Digital hard wired aids currently in the market have amplification characteristics that can be adjusted very flexibly.

General arithmetic processor:

This type of hearing aid has an arithmetic processor at its heart. This is like a computer that could perform activities as directed by the software. There is actually no fundamental limit to what these aids can do. All programmable hearing aids use this technology where software can be used to customize the hearing aid as per the requirement of the user.

The way the hearing aids amplify sounds greatly depends on the frequency. In order to achieve this frequency dependent amplification digital hearing aids process sounds in two different ways. The first one happens to be analogous to how analog hearing aids operate that is processing the incoming signal sequentially. An alternative or the second method happens to be block processing also known as frame processing or windowing the signal. In this approach, a number of input samples are taken in by the hearing aid before any computations on them are performed. Processing a complete block of input data at one time enables a Fourier transform to be calculated. Block processing enables complex calculations and operations to be performed.

Digital to analog converters

After the digital processing of the signal in a desired manner, the hearing aid must present the modified and amplified sound to the hearing aid wearer. Ears cannot decipher digital signals (signal presented as numbers) hence they must be converted back to acoustic signal and this is done by the digital-to-analog converter. The DAC is combined with the hearing aid receiver. This ensures that the digital signals are converted into analog voltage which is fed to a receiver that makes the final conversion of current into sound waves.

In order to minimize power consumption, digital hearing aids use the following method. The multiple bits that comprise each sample are converted into a single bit that changes at a rate many times higher than the sample rate. This converter is known as digital-to-digital converter. This high speed serial output from the converter is fed to the receiver which averages out the high speed variations in the digital signal thereby acting as a low pass filter producing a smooth analog signal. The digital-to-digital converter and the receiver combines to make up the digital to analog converter.

Specifications of digital hearing aids

These hearing aids have the same types of specifications like the gain, maximum output, range of frequency response adjustment, compression characteristics, internal noise and current consumption as is the case with analog counterparts. In digital hearing aids some additional specifications indicate the likely audio quality and processing capabilities significantly affect the sound quality and sophistication of processing these hearing aids provide.

Digital hearing aids can be characterized by the number of instructions or operations that they can perform in a second. A specific processor would be able to perform up to 40 MIPS (40 million instructions per second). Complex signal processing schemes require a greater number of instructions per second than less complex schemes. It should be stated that compression is more complex than peak clipping, multi-band processing is more complex than single band processing. If the processor within the digital hearing aid has a very high processing capacity then it will be associated with increased current consumption and would decrease the battery life. In an integrated circuit, increasing the number of instructions processed per second wound decrease battery life due to increased current consumption.

Sampling rate

Also known as sampling frequency, indicates how many times per second the hearing aid samples the input signal. Major impact of high sampling rate is that the hearing aid can amplify sounds only up to 40-45% of the sampling frequency with the absolute maximum being 50%. Another aspect of a very high sampling rate could be a limitation of complex sound processing by the hearing aid.

Number of bits

Each sample of audio waveform is represented by a number, which in turn is represented as a string of bits. The greater the number of bits, the greater the number of analog voltage levels that can be represented. Greater the number of bits, the better will be the digital approximation of the signal. More the number of bits it is better. The hearing aid will be able to handle a greater dynamic range of signals without adding a noise of its own.

Current consumption

The current consumption and hence the battery life and the feasible size of the battery depends on the instruction rate, the voltage at which the integrated circuit operates and the technology used to make the integrated circuit. The current consumption to power the digital processing for a given number of instructions per second is steadily decreasing.

Processing delay

Excessive time delay from input to output of the hearing aid will degrade signal quality. This is more so for the aid wearer's own voice. Longer delays facilitate more sophisticated signal processing that enables the hearing aid to react promptly and smoothly to changes in the signal dynamics.

Size

Complex circuits, and the power supply unit decides the size of the hearing aid. The size of the transducers are rapidly shrinking in size for the past 50 years there is a corresponding decrease in the size of the hearing aid.

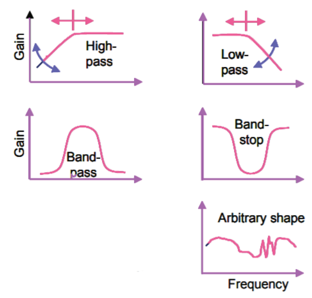


Image 1.19 showing various filters used in hearing aids

Filters. Tone controls and Filter structures

Tone controls and filters are very important in all hearing aids as they enable the amplification of different frequency regions.

Filters

The electronic component that causes gain to vary with frequency is known as the filter. Filters are known by their effect on signals. The types of filters include:

High pass filters - These filters provide more gain to high frequency sounds than to low frequency sounds thus giving the sound a treble or shrill quality.

Low pass filters - These filters provide more gain to low frequency sounds than to high frequency sounds thus giving the sound a muffled or boomy

quality.

Band pass filters - These filters provide more gain to frequencies in a certain band than to either higher or lower frequencies

Band stop filters - These filters provided less gain within a restricted range of frequencies than for all other frequencies.

All the filtering performed in hearing aids is achieved by mathematical manipulations while the signal is in digital form.

Tone controls

Function of tone controls in hearing aid is the same as in other audio devices. Tone controls get their name because they affect the tonal quality or timbre of sounds that pass through them.

Filter structures

Filters can be combined in serial, parallel or serial-parallel arrangements.

Serial structures - This arrangement comprises of one low pass and one high pass filter. This arrangement is called as serial structure, because all sounds pass through all the blocks one after the other. This arrangement is common in analog hearing aids, but is sparingly used these days.

Parallel structures - This arrangement generally allows more flexibility, even with simple filters. The filters divide the sound into adjacent frequency regions. These regions are also known as bands or channels. Parallel structures are simple conceptually speaking as the sound in each frequency region can be amplified more or less independently of sound in other regions. After the parts of the signal falling within each band have been amplified to the required degree and the parts are recombined. When the outputs from all the channels are recombined, the multiple versions of a single frequency component can recombine in a destructive manner at some frequencies and in a constructive manner at other frequencies thus imparting undesired ripples in the gain frequency response.

Serial-parallel structures - In parallel structures distortion is common. Serial-parallel structures can be used to avoid distortion.

Receivers

The receiver which resembles a microphone converts the amplified and modified electrical signal into to an acoustic output signal. The receiver operates by magnetic forces. Current passes through a coil that encloses a piece of metal, temporarily converting it into a magnet. As the current alternates in direction, this piece of metal also known as the armature, is alternately attracted and repelled by two permanent magnets. The free end of the armature is linked by a drive-pin to the diaphragm, so that the diaphragm also vibrates backwards and forwards and this produces the sound. Greater output can be obtained only by using a receiver with a bigger diaphragm which increases the size of the receiver or with the magnets further apart, which then requires greater electrical power for the receiver to operate.

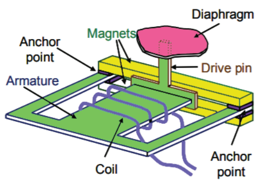


Image 1.20 showing the moving coil receiver

Frequency response of a behind the ear hearing aid has many bumps and dips. These bumps and dips are mostly caused by the tubing. The combined length of the tube in BTE model hearing aid has a length of 3 inches. The ear canal end of the tubing system connects to the receiver. Acoustically the tube has one end almost open and the other end closed. These tubes have a wavelength resonances at frequencies equal to odd multiples of the speed of sound divided by four times the length of the tube. This produces resonances at around 1 kHz, 3 kHz, and 5kHz. The resonance at 4 kHz appears to be due ot Helmholtz resonance between the mass of air in the tube and the volume of air inside the receiver. The bump at 2 kHz is primarily caused by the mechanical resonance of the receiver.

Acoustic dampers

Bumps and dips in the receiver response curve can adversely affect both the intelligibility of speech and quality of the amplified sound. Peaks become objectionable if they rise by more than 6 dB above the smooth curve joining the dips. Peaks caused by the receiver and tubing affect the shape of the maximum output curve of the hearing aid just as much as they affect the shape of the gain-frequency response. These peaks make it difficult to get all sounds loud enough without some sound becoming excessively loud.

Placement of an acoustic resistor (damper) in the tubing at the appropriate place decreases these peaks. One type of damper consists of a fine mesh like a fly screen is inserted across a small metal cylinder. As particles of air move backwards and forwards, in response to the sound wave in the tube they lose energy when they have to change course slightly to avoid the wires in the mesh. More quickly the particles are flowing, the more energy they will lose when the mesh is added. In a tube, the particles flow quickly

At resonant frequencies

At the open end of the tube

At any location a half-wavelength away from an open end.

A damper will decrease the receiver output most at the resonant frequencies, but this can happen only if the damper is placed in an appropriate place.

Dampers can also be made from sintered stainless steel (fine particles co metal). Another commonly used variety appears like a star shaped prism and hence known as the star damper. Dampers can also be made from lamb's wool and also from plastic. The degree to which a damper decreases resonant peaks depends on the impedance of the damper which in turn is determined by the fineness, length, and number of air paths through the damper. Fused mesh dampers and sintered-steel dampers are available in a range of standard impedances. The impedance of star dampers, lamb's wool dampers

and foam dampers can be varied by using different lengths of the material.

Dampers can be placed in the tubing connected to the receiver/in the inlet port of the microphone. Some receivers have dampers built in when they are manufactured.

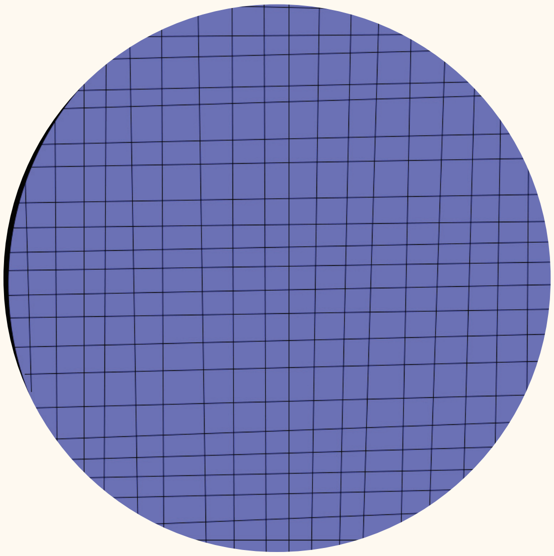


Image 1.21 showing Mesh damper

Telecoils

This is a small coil of wire that is capable of producing a voltage when an alternating magnetic field flows through it. The magnetic field that is picked up by the telecoil is usually generated by an electrical current that has the same waveform as the original audio signal. This magnetic field could occur as a by-product of some device, such as from a loudspeaker or a receiver in a telephone, or may be generated intentionally by a loop of wire around a room or other small area. The process of an electrical current inducing a voltage in a coil some distance away is known as the induction.

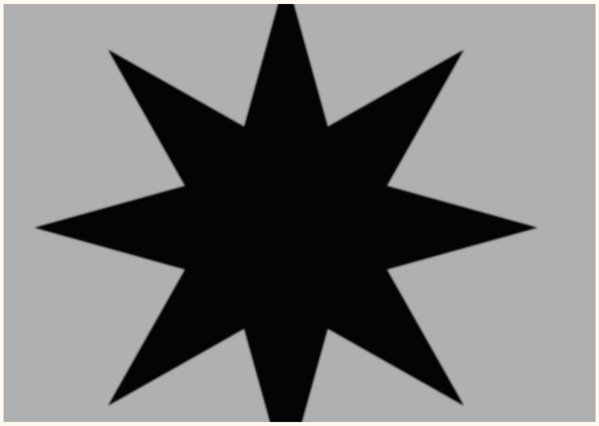


Image 1.22 showing Star damper

In order to improve the effectiveness of the telecoil the wire is coiled around a rod made of ferrite material as it provides a very easy path for magnetic fields to flow through. This material attracts and concentrates the magnetic flux. If more flux flows through the coil, then more voltage is generated by the coil. This is desirable because of the need to have a large audio signal compared to that of the internal noise generated by the hearing aid. The sensitivity of the coil can be increased by increasing the cross-sectional area or by increasing the number of turns. Both both these efforts increase the size of the coil and ultimately increase the size of the hearing aid.

It should be pointed out that not all hearing aids include a telecoil, although nearly all high-powered behind the ear hearing aids and many other hearing aids to have these coils. The user of the hearing aid can select the coil, instead of the microphone for amplification by switching the hearing aid to the T position. Many currently available hearing aids now have a program selector switch rather than a separate M-T switch. The telecoil can be selected by

switching to the telephone program. When other programs are selected then only the microphone is connected to the hearing aid amplifier. Some hearing aids can be set up to have a program in which the microphone and coil provide signals at the same time. This combination is also known as MT combination which is useful when the aid user wants to receive both acoustic and magnetic signals simultaneously or in quick succession. This setting has a disadvantage in that any acoustic noise present would also be amplified even if the aid user is trying to listen only to the magnetic signal.

When the hearing aid is switched to the T program and the input comes from a room loop then the only sound that is amplified will be the magnetic signals that reaches the hearing aid. When the hearing aid user uses the T program setting while speaking over the phone, then the microphone of the telephone will pick up any sound reaching it and the telephone side tone will cause all these sounds to emerge as magnetic signals and the hearing aid will sense these and amplify. In order to reduce the local noise while listening to the telephone conversation in the T program mode, the hearing aid user should cover the telephone microphone when not speaking through it.

Remote controls

They serve the same function for hearing aids as they do for television. This allows the user to vary the way in which the device works without actually touching it. This will be useful in managing really small hearing aids. It should also be pointed out that since some of the hearing aids are positioned behind the ear the user will not be able to see the controls. Remote control really helps to change the settings of the hearing aid without the need to actually touching them.

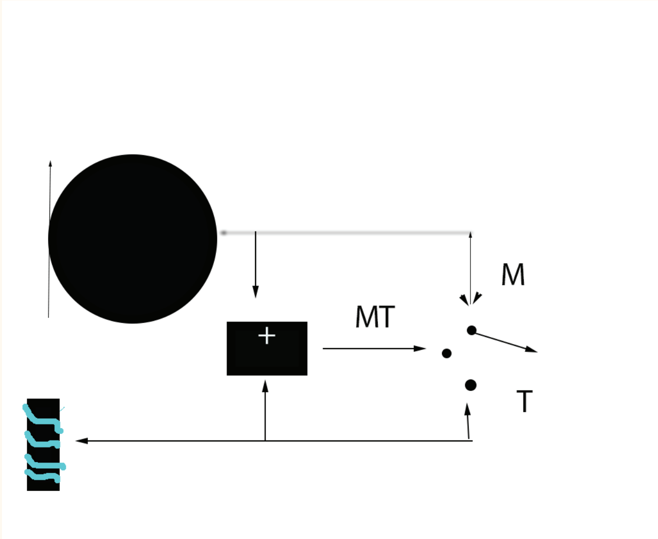


Image 1.23 showing diagrammatic representation of M and T settings

Buttons on the remote control are easier to operate than those on the hearing aid, partly because theey are larger and partly because the user can look directly at the controls while they are being operated. Some users would like to operate the remote control while keeping it within the pocket without attracting attention.

Remote controls work by transmitting signals to the hearing aid. These signals can have any effect on the hearing aid. Various methods of transmission are used in hearing aids. They include:

Infra-red:

This uses infrared technology the same as seen in television remote control devices. For this to work the remote control should be within the line of sight of the hearing aid. Ideally the remote should be pointed at the hearing aid for it to work. These hearing aids contains an infra-red detector on its

exterior. Bright sunlight could interfere with its function.

Ultrasonic:

This type of remote control transmits an acoustic signal that is too high in its frequency thereby making it inaudible by humans. These signals can be received by the hearing aid microphone. This remote also requires line-of-sight operation.

Radio wave:

An electromagnetic radio wave is transmitted by the remote and is received by a small aerial within the hearing aid.

Magnetic induction:

Control signals are transmitted from the remote to the hearing aid by creating a magnetic field at a frequency above the audible range. The hearing aid receives this using either a special purpose coil or the same telecoil that receives audio magnetic signals. Magnetic induction and radio wave controls have replaced ultrasonic and infrared remote controls.

Major concern with the use of remote control is their potential to interfere with pacemakers. The pacemakers are designed to sense small voltages, it is possible that the signals arising from the remote control could interfere with it.

Currently available pacemakers are not affected by remote control devices.
Bone conductors

These are alternative output transducer intended for people who cannot use the conventional hearing aid coupled to the ear canal. This can happen in external auditory canal atresia. Bone conductor transducers directly vibrate the skull and these vibrations are transmitted to the cochlea. The bone conductor works on the principle similar to that of a receiver. One major difference being that instead of a vibrating light diaphragm, it has a heavy mass that is shaken by the audio current passing through the coil. The inertia of this mass causes it to resist being shaken, so that the case of the vibrator shakes in the direction opposite to that of the mass. This vibration is transferred to the skull.

For efficient power transfer the transducer needs to be held firmly against the skull by means of a tight head band or spectacle frame. Continued wearing can create permanent indentation in the skull of the aid user. These devices require considerable power to provide the vibrations necessary for transferring sound directly to cochlea via bone.

The hearing aid amplifier output is connected to the bone conductor transducer, instead of its usual receiver by wires emerging from the hearing aid or by a plug and socket arrangement.

Image 1.24 showing bone conduction hearing aid

Batteries

Power source is very important for the functioning of hearing aid. It should be considered as an indispensable part of the hearing aid. Common cause of hearing aid failure happens to be due to drained out batteries. The important characteristics of battery are its voltage, its capacity, the maximum current it can supply and its electrical impedance. Another aspect that needs to be considered is its size.

Batteries generate electricity by putting two different materials (electrodes) in close proximity in a medium called the electrolyte that conducts electricity in the form of ions. Charged particles are attracted from one of the materials to the other via the electrolyte, and this can continue only if electrons can get from electrode to the other via an external electrical circuit. This external current of electrons is used by the hearing aid amplifier to power itself. This process continues till one of the electrodes is used up and it can no longer supply charged particles and electrons. This causes the battery to die out.

Voltage generated

Voltage generated by a battery depends on the type of materials used for the electrodes. Batteries commonly used in hearing aids use Zinc and oxygen as their negative and positive electrodes and hence these batteries are also known as zinc-air batteries. These batteries independent of its physical size can generate approximately about 1.4 volts when not connected to any device. It becomes 1.25 volts when it is connected to a device. When the zinc in the battery is close to being depleted, the battery voltage drops suddenly, and the hearing aid gets weak and the sound becomes distorted and could eventually cease. Some high end hearing aids could operate even when the voltage supplied by the battery drops below 1.0 volt. Some hearing aids could become unstable when the battery reaches the end of its life. They could make motor boating sound. Most of the hearing aids sense the voltage drop and would generate warning sounds indicating that the battery is about to die. Some hearing aids could reduce its function but producing lesser amplification as the voltage supplied by the battery undergoes a drop. This prevents sudden cessation of hearing aid function.

Body worn hearing aids use larger batteries like AA / AAA size. Some high end hearing aids use Lithium instead of zinc as the electrode. These batteries are capable of generating about 3 volts and are expensive.

Capacity and size of the battery:

Batteries last longer if the electrode material contained inside it is more. Bigger batteries hence last longer than smaller batteries with the same chemistry. The electrical capacity of a battery is measured in milliamp hours (9mAh). A battery with a capacity of 100 mAh can supply 0.5 mA for 200 hours, 1 mA for 100 hours or 2 mA for 50 hours.

High powered hearing aids need the maximum current to operate. Batteries used in these hearing aids belong to the High performance category and they are indicated by the prefix H in their name. These batteries are also zinc air cell type, but they have bigger holes to allow oxygen in at a faster rate and use an electrolyte that causes less voltage drop during high current demand. High performance batteries gives longer life than a standard battery. If the hearing aid draws more power then its life could be reduced.

Rechargeable batteries:

Some hearing aid manufacturers produce hearing aids with rechargeable batteries. Main advantage of these hearing aids is the increased convenience of not having to change the batteries. Rechargeable cells can be discharged and recharged several hundred times and the battery needs to be replaced only once in 2-3 years depending on the usage.

Major drawback of rechargeable cells is that their capacity is only around 10% of that of a non-rechargeable cell of the same size and recharging should be performed on a regular basis. In many hearing aids rechargeable cells will not be able to provide sufficient current when the wireless feature of the hearing aid is enabled.

Rechargeable cells used in hearing aids are commonly of Nickel metal hydride construction. They generate 1.2 V which remains constant as the cell is discharged. A potential advantage of rechargeable batteries is that they can be recharged from a solar cell, making their use suitable in places where there is a reliable supply of sunlight.

Batteries some practical tips:

1. Sticky tabs present on zinc air batteries restricts air from getting into contact with electrodes. The battery will not function till the tab is removed. Once this tab is removed the battery becomes functional and its shelf life is restricted to only a few weeks.

2. If the sticky tab is too well sealed the battery, then it will not be usable until the air has had time to percolate into the battery. This process can be speeded up by leaving the battery a few minutes before inserting them into the hearing aid.

3. If a new battery appears dead, then it can be left for a few minutes. It could miraculously start its function.

4. If a hearing aid is left unused for a period lasting weeks and months then the battery should be removed to protect the hearing aid from potential battery leakage and corrosion.

5. If high powered hearing aid is used then it is worth investigating the battery life and sound quality obtainable with a high performance battery.

This chapter will describe the process of combining bits and pieces of electronic circuitry to construct an hearing aid that could be used effectively to rehabilitate persons with hearing disability.

Types of hearing aids:

Body worn hearing aid:
These aids were the first portable electronic hearing aids. This type of hearing aid was designed by Harvey Fletcher while working at Bell Laboratories. Body worn hearing aids consists of a case and an ear mold that is attached by a wire. The case worn in the body / pocket of the person contains the electronic amplifier components, controls and battery. The ear mold typically contains a miniature Loudspeaker.

Image 1.25 showing a body worn hearing aid

The case typically is about the size of a pack of playing cards and can easily fit inside shirt pocket or on a belt. Major advantage of these hearing aids is that these aids can provide greater amplification with a longer battery life and costs significantly less than other types of hearing aids.

Behind the ear hearing aids:

These hearing aids are worn behind the ear. These hearing aids are available in two different classes.
1. Behind the ear
2. In the ear

Behind the ear model is characterized by the case that hangs behind the pinna along the post aural groove. This casing is attached to the ear mold or dome tip by a tube which is made of plastic / silicone. This tube crosses from the superior-ventral portion of the pinna to the concha, where the earmold or dome tip inserts into the external auditory canal. The case contains the electronics, controls, battery and a microphone. It may even contain multiple microphones. The loudspeaker / receiver may be housed within the case as in the case of traditional behind the ear model.
The speaker / receiver is placed within the ear mold / dome tip in the case of receiver in the ear canal.

This type i.e. receiver in the ear canal is smaller than the traditional behind the ear model and is commonly used in active population. Behind the ear hearing aids can produce more output and could hence be used to manage patients with more severe degrees of hearing loss. It is so versatile that it can be used to manage nearly all types of hearing losses. These aids are portable and easy to handle. These devices can easily be connected to assistive listening devices like FM / induction loop

systems. Behind the ear hearing aids are commonly used in children who need a durable type of hearing aid.

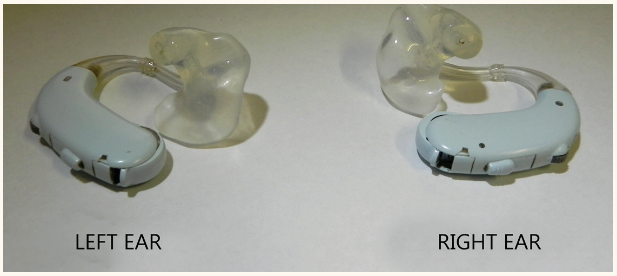


Image 1.26 showing behind the ear model hearing aid

The configuration i.e. mirrors if these aids need to be used for both ears as shown in the figure above. Whereas in the body worn hearing aid the same unit can be used to connect to both ears by providing two tubes connected to the same box using a splitter.

In the ear hearing aids:

These devices fit in the outer ear bowl otherwise known as the concha of the pinna. Since these aids are larger than that of intracanalicular hearing aids it can hold extra features. These aids should be custom built to fit into each individual ears.

These aids could sometimes be visible when standing face to face with the user. These aids can be used to manage patients with mild to moderate levels of hearing loss.

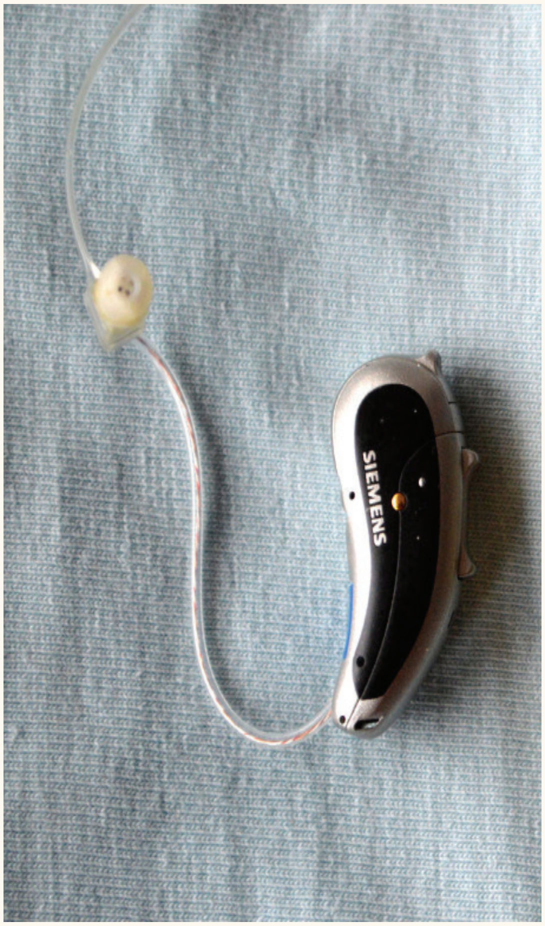


Image 1.27 showing receiver in the canal hearing aid

Feedback (squealing / whistling sound) could be a major problem with these hearing aids. Some advanced models provide circuits that could cancel out the feedback. Venting can increase the feedback. A vent is actually a tube placed to offer pressure equalization within the ear canal. Different vent sizes have been used in an effort to prevent this feed back effect.

This type of hearing aid is not recommended for children as they could find it difficult to retain them. Moreover the mold needs to be replaced periodically as the size of the concha increases due to growth of pinna in children.

This type of hearing aids can be connected wirelessly to FM systems which is provided within the classroom of these persons.

Mini in canal (MIC) or completely in canal (CIC) hearing aids:
These hearing aids are generally not visible unless the viewer looks directly into the ear canal. These hearing aids are intended for persons with mild to moderate hearing loss. Completely inside the canal hearing aids are not recommended for persons with good low frequency hearing as the occlusion effect could make these sounds more noticeable and make them irksome.

The completely in the canal hearing aid fit tightly deep in the external auditory canal. It is hence barely visible. Since it is small it will not be provided with a directional microphone and its exceedingly small batteries have a short life. The controls of these aids will be difficult to access by old age persons.

These aids help the wearer during telephone conversation. There is also no feedback. In the canal hearing aids are also placed deep within the external auditory canal. They are barely visible. Larger versions of these hearing aids could be provided with directional microphones. These aids are relatively easy to manipulate as they are placed superficially within the ear canal.

These hearing aids are relatively more expensive than their behind the ear counter parts. Miniaturization increases the cost. More the miniaturization more becomes the cost. Ear molds need to customized to fit into the ear canal for which use of CAD becomes necessary.

Invisible in canal hearing aids:

These hearing aids fit inside the external auditory canal completely. The device is completely not visible from the outside. The fit is also comfortable as it is custom made to suit the external auditory canal of the individual. These hearing aids use the principle of venting to equalize external auditory canal pressure. These models enable the user to use mobile phone as a remote control to alter memory and volume settings. This type of hearing aids are not suitable for old individuals because they lack the dexterity to handle these devices.

Extended wear hearing aid:

These hearing aids are placed deep inside the ear canal and can be worn for several months at a time without removal. This type of hearing aid is suited for persons with moderate to severe hearing loss.

These devices provide reduced distortion, reduced wind noise, and reduced feedback. It also provides directionality and better quality when compared to other hearing aids.

These hearing aids are made of soft material and is designed to fit the contour of the external auditory canal. These hearing aids are inserted non surgically by an otolaryngologist / audiologist. This device is placed within the bony portion of the external auditory canal just 4 mm away from the ear drum. This device needs to be removed if battery goes dead in order to replace the battery.

Image 1.28 showing in the canal hearing aid. Blue arrow shows the vent

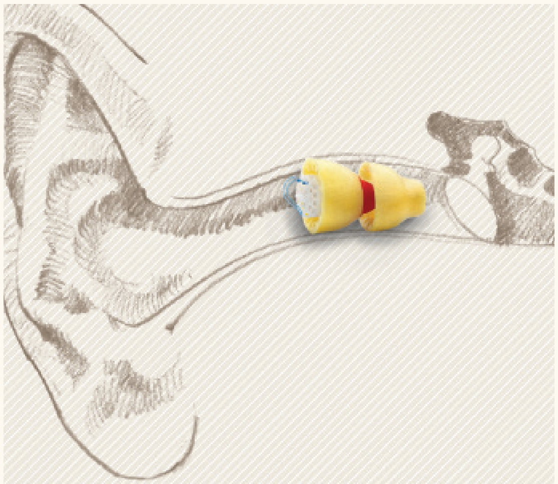


Image 1.29 showing completely inside the canal hearing aid

Extended wear hearing aid:

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Spectacle hearing aids:

These hearing aids are built into the frame of the spectacles and can be used by individuals wearing glasses. This hearing aid is just used for cosmetic effects. There are two types of Spectacle hearing aids. They are bone conduction spectacles and air conduction spectacles.

Bone conduction spectacles:

Sounds are transmitted via a receiver attached to the arm of the spectacles which are fitted firmly behind the bony portion of the skull at the back of the ear (mastoid process). The sound passes from the receiver on the arm of the spectacles to the inner ear via bone conduction mechanism. This type of sound conduction requires more power than other types of hearing aids. These hearing aids give

poor high pitch response and are best suited for conductive hearing loss or where it is impractical to fit standard hearing aids.

Air conduction spectacles:

In this type the sound is transmitted via air conduction which is facilitated by a small rubber tube attached to the arm of the spectacles. The tube is connected to the ear mold.

Custom hearing aids

In the canal hearing aid and completely inside the canal hearing aids can be custom built for individual hearing aid user. Similarly any of these styles can be manufactured in totally standardized shapes and sizes, this is referred to as a modular construction. Hearing aids can also be constructed in a semi custom / semi modular manner.

Custom hearing aids (ITE's, ITC's and CIC's) take full advantage of the size and shape of the individual user's ear contour.

Construction of custom hearing aid usually begins when the clinician makes an ear impression and sends it to the hearing aid manufacturer. A laser scanned image of the ear impression can also be sent to the hearing aid manufacturer these days. The manufacturer uses the impression to create a hollow shell that could fit snuggly into the external auditory canal of the user or sometimes an aid that could fit into the well of the concha can be designed.

Customization of hearing aid components is carried out in varying degrees. The basic construction of the hearing aid begins around the faceplate. Faceplate is a flat sheet of plastic that is trimmed to size and becomes the outer surface of the hearing aid. This faceplate will come from the factory with thee amplifier board, microphone, volume control, battery compartment, telecoil and switch already assembled. The receiver would be loosely attached to the remaining components, because its position relative to other components has to be adjusted for each custom hearing aid to make the best use of the space available within each ear.

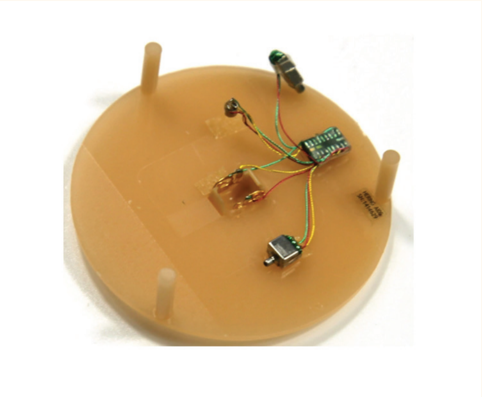


Image 1.30 showing a typical faceplate the starting point for hearing aid construction

In some kits one or more components may not be positioned. This provides scope for further customization of the hearing aid. This also ensures that optimal use of the space. As much material as possible can be cut from the outer part of the plastic shell and the hearing aid can be shrunken still further. The faceplate is finally glued to the shell and any excess of the plate can be trimmed off. The miniaturization causes problems during repairs of hearing aid. Most of the repairs need to be executed after cutting the faceplate away from the shell. This can in majority of cases can be done, but still it remains a tricky process.

Modular hearing aids

In these hearing aids the components are manufactured as a standard package. ITC hearing aid (in the canal) have been manufactured in a variety of shapes (standard). They can be purchased as ready to wear hearing aids. Physical fitting of these hearing aid require selecting the case with the shape that best matches the user's ear canal and incorporating the modular hearing aid components within it.

Modular hearing aids has its own advantage and disadvantages. Major advantage being that the module can be manufactured and tested in an automated manner which lowers the cost of production and improves reliability. It also provides the attractive option of the ability to fit the hearing aid as soon as the hearing assessment is completed.

The main disadvantage is that there is no standard casing available that could fit into all human ear canals in a cosmetically and functionally acceptable manner. There is also the added risk of the hearing aid falling off from the ear canal. If there is sound leakage around the casing then there could be intense whistling and feed back which could be irksome to the patient. Low cost hearing aids are modular devices. These aids have a sleeve made of foam around the canal section of the aid. This in some way negates the problem of loose fit and feed back as a result of loose fit. The basic disadvantage of this foam is that it could deteriorate with time and needs to be replaced.

Disposable ITC hearing aid

This is another type of modular hearing aid. This was succeeded by a disposable BTE (behind the ear) hearing aid. These aids contain an embedded battery and the complete hearing aid must bee replaced when the battery gets fully exhausted. These devices proved to be extremely cheap and it is combined with good quality electronics and transducers. The ITC type of disposable aids are available in a range of gain-frequency response shapes. Their frequency response, internal noise, and distortion compared favorably with more expensive hearing aids.

BTE and body level hearing aids could also be considered under this modular category. The electronics and mechanical components of these aids have a fixed size and shape. They are connected to either an individual earmold or a standard dome located in the ear canal. When used with a standard dome, either as a tube fitting or with a receiver in the ear canal(RITE) BTE's are completely modular.

Semi-modular, Semi-custom hearing aids

ITE / ITC hearing aids that combine a standard module with that of a custom made ear shell can be classified under this category. The modules are usually clipped, rather than glued to the individual ear shell. This ensures that the repair process is faster, cheaper and the risk of damage to the earshell or faceplate is minimized. Major disadvantage of these hearing aids is that the components cannot be rearranged to take advantage of the individual ear's geometry. A semimodular hearing aid would be generally larger than a custom hearing aid with the same components.

Reliability of hearing aids

Hearing aids need to function in adverse environment like exposure to rain, sweat, cerumen, hairsprays, hair gels and humidity. All these factors either individually or collectively could cause the hearing aid to fail. The components that commonly fail are those exposed to air or those parts that are

moving. They include battery contacts, transducers, volume controls and switches.

Some of the following innovations has greatly reduced the failure rate of hearing aids.

1. Automatic volume control - This includes a wide dynamic range compression thus making it unnecessary for hearing aids to incorporate a manual volume control.

2. Electrical programming - This feature reduces the number of components with moving parts.

3. Water repellent fabric - This type of fabric covers the microphone inlet port, impedes the entry of water into the microphone.

4. Water proof membranes - They block the entry of moisture and cerumen into the receiver, but allows sound waves to pass through unimpeded.

5. Gaskets and water proof fabric - This feature enables the zinc air battery to obtain the oxygen supply it needs to operate without allowing water into the battery compartment.

6. Nanocoating - A lacquer containing nano particles makes the surface so smooth that water beads on the surface instead of spreading over it making the water entry less likely.

7. Swipe control - They sense the movement of the finger, and don't require any moving or mechanical parts or apertures that enables ingress of moisture.

Linked bilateral hearing aids

It has been a custom for many years to fit a hearing aid to both ears. Currently technology has made it easy for these two hearing aids to exchange information with each other so that they can perform their function in a co-ordinated manner. This feature is desirable for the sake of convenience and performance.

Let us consider the following scenario: A hearing aid user who wishes to adjust the volume. It will be cumbersome to adjust the volume of both hearing aids and it could mean sense if the volume is raised in one ear and the same is carried out to the other ear. Since the second hearing aid does not need its own volume control the freed up space can be used for a program selector switch.

The performance advantages of linked hearing aids are not that marked. Ensuring that the two hearing aids make the same choice of microphone directivity at any instant minimizes the likelihood that the hearing aids will distort the timing and level differences between the ears that the user uses to localize sounds. co-ordinating compression and adaptive noise suppression in the two aids minimizes the distortion of inter-aural level differences that will occur otherwise. Linked processing appears to slightly improve localization and naturalness of the sound heard.

Linked bilateral hearing aids can also make decisions about when to switch he input from the microphone to telecoil. A single hearing aid cannot make a correct decision of switching on to a telecoil from microphone mode when the user walks past some piece of equipment that generates a strong magnetic signal. Similarly telephone held to one ear will ensure that there is a stronger magnetic field occurs closest to the hearing aid than in the distant one. The hearing aid nearest to the telephone receiver can switch to telecoil mode while the aid in the other ear is in the microphone mode.

Many linked bilateral hearing aids can exchange control information such as volume settings, program settings, directional microphone settings and information about how much gain a compressor is currently providing. The control signals between the hearing aids have a very slow rate of change and hence information can be transmitted with a very low radio frequency bandwidth. This requires very little power.

Some hearing aids can transmit a full audio bandwidth signal from ear to ear. In some hearing aids this is achieved by near-field magnetic inductive coupling. Irrespective of the technique used, if a full audio bandwidth signal can be transmitted from one ear to the other, signals sensed on one side of the head can be transferred to the hearing aid on the other side of the head enabling scope for several uses like:

1. Transferring the microphone signal on one side to the ear on the other side after amplification makes a wireless CROS hearing aid.

2. Unlike the conventional CROS hearing aid the transfer can be in either direction, and the direction can change from time to time depending on where the dominant speaker is situated. This feature is useful while traveling in a car where the sitting positions are fixed and conversation partners are limited. The signal to noise ration is much larger on one side of the head than the other.

3. Transferring a telecoil signal from one ear to the other, enables a telephone call to be heard in both ears simultaneously.

4. Signals from microphones on both sides of the head can be combined in each hearing aid to produce a greater degree of directivity to the front, or in any other direction.

Programmable hearing aids

Audiologists change the contents of the digital control circuits using a programming device. Hearing aids currently are programmed using computers. Initially special purpose programing devices were used for this purpose, but they have largely been replaced by computers. Currently all manufacturers of hearing aids have adopted a common standard for storing data and sending information from computers to hearing aids. This standard is known as the NOAH (to indicate we are all in the same boat).

The NOAH standard specifies how common data like the audiogram should be sent to and received from the hearing aid. In order to program hearing aids from different manufacturers, specific software provided by that manufacturer is needed.

Once the client's data has been entered, the same can be accessed from any manufacturer's program, so that potential fittings from different manufacturers can be compared as well. This feature is really exciting. Before a hearing aid can actually be programmed using a computer, it needs an interface which should be used to connect to the computer. This interface is a small box with suitable sockets and some circuitry incorporated into it. This interface goes by the name HiPro (hearing instrument programmer) interface. The HiPro device is still commonly used. Other alternatives are also available.

NOAH link wireless interface - This connects via a cable to the hearing aid exactly in the same way as HiPro, but has a Bluetooth wireless connection to the computer. When the NOAHlink is worn around the neck of the patient, he/ she can move freely in an unrestricted manner while the hearing aid is being programmed as long as they stay within the wireless range of the computer which is about

10 meters. Fine tuning of the hearing aid can be done in any environment if the clinician uses a laptop computer.

The NOAH-link wireless interface can plug into the nEARCom, a hook shaped device worn around the patient's neck. This device contains a 10.6 MHz inductive transmitter/receiver module to send signals to and from the hearing aid. The transmitter-receiver modules are manufacturer specific, but up to five of them can be inserted in the nEARCom at the same time.

Multi-memory / Multi-program hearing aids

Data sent to the hearing aid by the computer is stored in a memory inside the hearing aid. If one set of data can give the hearing aid one set of performance characteristics (e.g.. gain, frequency response, microphone directionality) then several sets of data can give the hearing aid several sets of performance characteristics. Each set of performance characteristics is called a hearing aid program. Either the user or the hearing aid itself can switch between programs whenever appropriate.

The reason behind the user's desire to change the program is that sounds entering the hearing aid can have acoustic properties that differ greatly from one environment to another. For optimal listening, the hearing aid should have different amplification characteristics in each environment. A programmed hearing aid can sense the acoustic environment and can automatically change the amplification characteristics. It is however possible that the user can do a better job in selecting the optimal characteristics than an automatic circuit.

In majority of multi-memory hearing aids, all of the parameters that can be adjusted in one program can be independently adjusted in the other program or programs.

Paired comparisons

If hearing aids are given two or more programs, then they can rapidly be switched between two programs during the fitting process. This enables the hearing aid wearer to compare two responses in quick succession and state which one is preferable. The audiologist can use these preferences to fine-tune the response when the hearing aid is initially programmed and at any follow up appointments.

Remote sensing and transmitting hearing aid systems

When sound wave travels away from its source, its power spreads out over an ever increasing area and hence it gets weaker. This causes two types of sound quality degradation. First, the decreased level is more easily masked by the background noise. Secondly, reflected sounds in the form of reverberation, add delayed versions of the original sound to the direct sound. Reverberant sound is smeared out in time and is much less intelligible than direct sound, particularly so when the room has a long reverberation time. Noise and reverberation both cause intelligibility to diminish as the listener gets further from the source of sound.

In light of the above facts the term critical distance is rather important. Critical distance is defined as the distance from the source at which the level of reverberant sound equals the level of the direct sound. Beyond this critical distance the reverberant sound dominates. Larger the room, and the less reverberant it is, the greater is the critical distance. In classrooms the critical distance is often in the range of 1-2m, and in a living room it is typically a little less than 1 m.

A solution to this problem of reverberation masking the direct signal is to pick up the signal where it is strongest and clearest (i.e. next to the speaker's mouth) and transmit this strong signal which is clear to the hearing aid wearer either as an electromagnetic wave or as a magnetic field rather than as a sound wave. Of course the hearing aid user should have the equipment necessary to turn the electromagnetic wave back to sound wave. There are three types of wireless transmission systems used to get the signal from the talker to the listener. They include:

Induction loops:

There is a close connection between electricity and magnetism. Induction loops take advantage of this by converting an audio signal into an electric current that flows through a loop of wire and hence into a magnetic field that ravels across space at the speed of light. This magnetic field is sensed by a coil of wire, and it induces an electrical voltage in the coil. This voltage is then amplified and converted by a receiver back into sound. The loop that emits the magnetic field can be as large as a length of wire around the perimeter of an auditorium or as small as a device that can fit behind the ear alongside the usual hearing aid. The coil that picks up the magnetic signal is invariably mounted inside the hearing aid used by the listener.

Field uniformity and direction

Magnetic fields actually flow in circles around the current that induces the field. As these circles become more distant from the current, the magnetic force and the magnetic flux become weaker. To visualize the flow of magnetism, angle the thumb at right angles to the fingers of the right hand and curl the fingers. If the thumb points in the direction of the electrical current, the curled fingers will show the circular path taken by the magnetic field around the line of the thumb. Engineers call this right hand rule and use it to deduce which way around the circle the magnetism is flowing.

In order to work optimally with room loops, the telecoil in a hearing aid is mounted vertically and will pick up only the vertical part of the magnetic field. Since the source of magnetic signal is sound the circular magnetic fields will reverse their direction many times per second. It is this changing magnetic flux that enables a telecoil to sense the magnetism and produce an audio voltage. It should be pointed out that the earth's magnetic field does not induce a voltage in the coil, because the earth's field has a constant strength and direction.

Magnetic field strength

The strength of the magnetic field near the center of the room is directly proportional to the magnitude of the current in the loop and to the number of turns in the loop and is inversely proportional to the diameter to the loop. It should be stressed that there are other sources of magnetic field within the room. It could also arise from electrical wires. Electrical wires produce a magnetic field ranging between 50-60 Hz depending on he country. This could be termed as magnetic interference (background noise) which has a characteristic hum or buzz. This frequency can easily be dampened inside the hearing aid.

Considering the standards for room loops and telephones, a hearing aid wearer should be able to conveniently switch from microphone to telecoil mode without changing the volume control if the hearing aid produces the same output for a magnetic field strength of 60-100 mA/m as it does for an acoustic input of around 65 dB SPL.

Loop frequency response

The frequency response of a loop and telecoil system can sometimes be unsatisfactory. The hearing aid acoustic response would have been carefully adjusted to suit the aid wearer and it is important that the combined response of the loop and hearing aid coil not be too different from the acoustic response. Once exception to this rule would be an additional cut for frequencies below about 500 Hz may be beneficial as this the the frequency region where magnetic interference is likely to occur.

Multi-memory hearing aids make it possible to adjust the response separately for the telecoil and microphone operation so that the best telecoil response for an individual aid user can be selected. Some remote control devices allows the user to select a low-tone cut when needed, such as in rooms with lot of magnetic interference. Adaptive noise reduction algorithms should also be effective at decreasing the buzz from magnetic interference because of the constant / slowly changing nature of the magnetic interference.

Possible reasons for the user to experience a different frequency response in the telecoil position than in the microphone position:

The loop may emit a weaker magnetic signal for high frequency sounds than it does for low frequency sounds. This could happen because the electrical impedance of the loop comprises of an inductance as well as resistance features. An inductance has an impedance that increases with frequency, so the total impedance of the loop starts to rise once the frequency exceeds a certain limit which is known as the corner frequency. At the corner frequency, the impedance of the inductance equals the impedance of the resistance. If the loop is powered by a conventional audio power amplifier, the current, and hence the magnetic signal, will both decrease as frequency rises above the corner frequency. The solution is to make sure that the corner frequency is 5 kHz or higher. This can be achieved by using:

1. Wire with a small diameter (provided it does not overheat)

2. A special current-drive power amplifier with a high output impedance

3. Installing several loops to cover the total area

4. Very few turns, or even just one turn, in the loop

5. A graphic equalizer

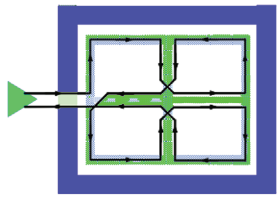
6. An external series resistor.

The last three options could require a more powerful amplifier.

Another alternative solution to this problem is to use a grid of small loops, that could be placed under the carpet or mat, rather than around the perimeter of the room. This design also minimizes spillover from the loop to adjacent rooms.

Another effective solution is to use two different loop systems in complex patterns covering the same area. The second loop is driven by a second amplifier, which produces a signal 90 degrees out of phase with the signal from the first amplifier. Each complex loop provides a field in the dead spots of the other loop, and the 90 degree phase shift prevents the two magnetic signals from canceling each other in places where they both are strong. This combination is known as the phased-array-loop, results in a very uniform magentic

field, even if the area is large, and minimal spill-over outside the area of the loops.



Multiloop connection covering a large area

Image 1.31 showing a multi-loop system covering a large area. Field strength is weak within the green shaded area.

Another reason why the telecoil frequency response might be different from the acoustic response lies within the hearing aid itself. Coils inherently produce a voltage that rises with frequency. Thee hearing aid designer could compensate for this either partially or completely by the way in which the telecoil connects to the hearing aid amplifier. The end result would likely to have a low frequency roll-off.

It is highly desirable for room loops to include a bass boost to compensate for the low cuts that usually occur in hearing aid telecoil circuits. This boost would also improve the ratio of signal levels to interference, which is usually very low frequency dominated.

Tips for wearing a transmitter and microphone:

1. Wearing the microphone on a head-mounted boom just below the mouth will result in a SNR about 10 dB better than clipping it to the lapel or dangling around the neck. The signal transmitted will not be affected by extreme head turns away from the microphone.

2. Clipping the transmitter microphone to the lapel will result in a SNR about 10 dB better than clipping it at waist level.

3. A directional microphone on the transmitter can compensate for it being worn further from the mouth

4. Wearing a transmitter with self-contained microphone under clothing, though convenient, is likely to produce clothing rubbing noises for the recipient as well as an attenuated signal which should be avoided.

Radio-frequency transmission

This provides a portable and reliable way to get the signal from a talker to a listener without corrupting the signal by noise or reverberation. The talker wears a small transmitter which contains a microphone, in which case it is worn around the neck. Alternatively a microphone can be attached to the transmitter cable, in which case the transmitter is clipped to a belt or worn in a pocket and the microphone is clipped to the lapel or worn on the head. The connecting cable also serves as an aerial for the transmitter.

The receiver is worn by the hearing impaired listener.

This device can be connected electrically to the

hearing aid by a cable (direct audio input).

It can connect by magnetic induction (via a loop around the neck or by a silhouette coil) to the hearing aid.

It can be clipped onto the hearing aid

The whole receiver can be incorporated within the hearing aid.

In radio-frequency transmission, the audio electrical signal is not directly converted to another form of energy as occurs in magnetic induction form a loop to a telecoil, but it modifies or modulates the electromagnetic wave. In the absence of an audio signal, the carrier resembles a sinusoidal wave. It can convey information only when the audio signal alters some aspect of the carrier. A variety of analog or digital modulation techniques can be used. The two most commonly associated with short range transmission are frequency modulation and frequency hopping spread-spectrum modulation.

Frequency modulation

Also known as the FM. It the hearing aid field, it is the carrier frequency that is commonly altered. The job of the receiver is to detect the carrier and then produce a voltage that is a replica of the original audio signal. The extraction of the modulating waveform is called demodulation.

Other forms of modulation are also available. Common of them include the amplitude modulation in which the audio signal modulates the amplitude rather than the frequency of the carrier.

The advantage of using modulation is that the strength of the audio signal coming out of the receiver is not dependent on the strength of the carrier wave and hence does not depend on the distance between the transmitter and receiver. As the carrier wave becomes weaker, the receiver will however add noise to the audio signal. When the carrier becomes extremely weak, reception will cease entirely.

In order to select the particular wave from many hundreds of electromagnetic waves by the process of tuning. When the frequency of the electromagnetic wave matches that of the frequency tuned then the receiver will pick up the transmitted signal.

In FM transmission and reception an additional phenomenon helps when the receiver is exposed to two different transmissions at the same carrier frequency or two carrier frequencies that are only slightly different. The demodulator works by locking on to the carrier and then measuring how much its frequency varies with time. It can lock on to a strong carrier even if a weaker carrier is simultaneously present. This phenomenon of demodulating only the stronger signal is known as the FM capture effect because the receiver is captured by the stronger signal. If two transmitters of the same output power are generating two signals then the stronger of the two signals at the receiver will be the one that originates from the closer of the two transmitters.

The field intensity coming from a transmitter decreases in inverse proportion to the square of the distance from the transmitter (inverse square law). Radio-frequency waves pass through non-conductive obstacles extremely well. They are attenuated by large conductors such as a sheet of metal and to a lesser extent by the human body. Large metal objects can cause signals arriving from distant transmitters to be received with stronger than expected intensity.

Metal objects can also cause reflections of the electromagnetic wave. This reflection can cancel the signal coming directly from the transmitter, thus

causing the signal strength to be very low at certain places in the room. The receiver at these positions will not be able to adequately detect the carrier and a dropout occurs. The listener will hear only the noise. Many advanced receivers will detect that a dropout has occurred and will mute or squelch he output signal, so that silence occurs when the receiver detects that it is not receiving the carrier wave.

Because it could be possible for the signal strength to be higher or lower than expected, a good solution if many classes are situated nearby is to provide each classroom with difference carrier frequencies to prevent confusions occurring.

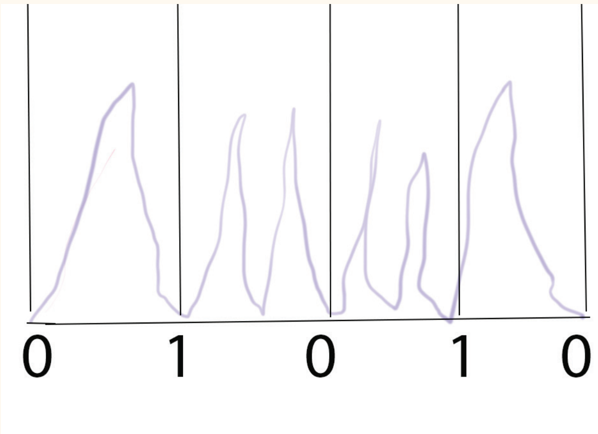


Image 1.32 Transmission of digital data by differential binary phase-shift keying

Digital modulation techniques

Alternative modulation techniques well suited to the transmission of digital data including digital audio is available. It is the combination of differential binary phase-shift keying and frequency-hopping spread spectrum.

In binary phase-shift keying, the 1's and 0's of the digital data are represented by the phase of the radio-frequency carrier. As an example, every time a digital 1 occurs, the phase of the carrier is changed by 180 degrees, which simply indicates inversion of the signal.

Every time a digital 0 occurs, the phase is left unchanged. A receiver can recover the digital data by detecting whenever the phase of the carrier changes. This method has the same disadvantage as FM, in that another transmitter sending the same frequency can interfere with reception.

Frequency hopping is the alternative that is less prone to interference. The transmitter hops to a new carrier frequency many times per second. This occurs in a pre-arranged but seemingly random sequence. If the receiver is aware of this hopping frequency, it also hops at the same time and the transmission of information carries on in an uninterrupted manner. At each new carrier frequency, the receiver continues to detect changes in the phase of the carrier, and so continues to recover the digital data. The advantage of this method being that the transmitter sends a very small amount of power at each frequency, and this signal is less likely to interfere with other transmitters, or be interfered by them. In order to keep the amount of information as low as possible, the bit rate of the digitalized audio signal is reduced by one of the several available data compression algorithms. In the receiver, the encoded signal is decoded to reconstruct a close approximation of the original signal.

Frequency hopping spread spectrum

Bluetooth is the most well known example of this type of modulation. Transmission occurs by hopping between 79 channels, each 1 MHz wide, in the range from 2402 MHz to 2480 MHz. Hops occur 1600 times per second, and if interference occurs in any carrier frequency, the transmitter and receiver agree to skip that frequency in the future, thus decreasing interference to and from conventional narrow-bandwidth transmitters. This is called as adaptive frequency hopping as the sequence adapts to avoid interference. Bluetooth systems are intended for short-range transmission, ideally 30 ft or 10 m and have many applications in audiology.

Bluetooth transmitter consume too much battery energy and hence it is difficult to include these inside the hearing aid because of energy constraints. The high frequency of operation (2400 MHz or 2.4 GHz) however enables operation with a very small antennae. Purpose designed frequency hopping systems operating in this frequency range are being incorporated within hearing aids.

Another disadvantage of Bluetooth, for some of the applications is the need for handshake transmission protocol which causes delay to audio signals. The sound is desynchronized from visual input, perhaps to such an extent disturbs lip-reading.

A major source of interference could occur if the user can hear the original sound source, plus the delayed version arriving via Bluetooth transmission. The Bluetooth delay creates no problem if there is no visual information and no audio information reaching the user except that which arrives via the Bluetooth device.

Recently a new low power Bluetooth standard that uses less power and has less audio delay has been introduced. So future hearing aids can make use of this device.

Coupling of ear to hearing aid

Audio signal coming from a wireless receiver would be useful only if it can be delivered to the ears of the user. Simplest form of this coupling of signal with the human ear is by directly driving an earphone which is fitted to the external auditory canal. Major disadvantage of this process is that wireless receivers don't usually contain sophisticated tone controls or adjustable forms of compression. Hence it is not possible to adjust the amplification characteristics to suit the requirements of the individual aid wearer.

Individual amplification needs can be met accurately if the wireless output is coupled to the person's own hearing aid. This coupling can be achieved by the following ways:

1. Electrically, from the body-worn receiver via a cable to the hearing aid's direct audio input connector provided it has one.

2. By the process of induction from the body-worn receiver via a loop worn around the user's neck, which sends a magnetic signal to the hearing aid telecoil or via a small coil mounted in thin plastic case that is positioned behind the wearer's ear, right beside the wearer's own BTE hearing aid. This coil is known as a silhouette coil, because its case has a profile similar to that of the BTE hearing aid. It is also known as an inductive ear-hook.

3. Electrically from a receiver mounted into a small boot that plugs into the bottom of the BTE hearing aid

4. Electrically, from a receiver completely integrated inside the hearing aid

It should be stated that each of these methods has their own advantages and disadvantages, and could affect the gain-frequency response of the combined amplification system in different ways. Direct electrical connections provide a well-defined signal. Since a cable needs to be used for the purpose of integration, it could be cosmetically undesirable, and less reliable after continued usage. Another unique advantage is that speech-operated switching and adaptive systems are most easily possible with direct connection. Stereo listening is possible if the source produces a stereo signal.

If a body-level receiver is used, the neck loop is cosmetically superior and there are no cables outside the clothing thereby making it acceptable. One major disadvantage is that low frequencies can be attenuated, and the strength of the magnetic coupling (hence the audio signal) can be decreased when the head is inclined to either side. Magnetic signals are more prone to interference from nearby electrical apparatus. The silhouette coil has all the disadvantages associated with the presence of a cable, and the potential for interference. Stereo connection is possible and unlike the neck loop there is no change in the signal strength with changes in the head position.

The coupling method that has the best combination of reliability and cosmetic acceptability is when the wireless receiver is mounted inside the hearing aid, or is mounted in a small boot that plugs onto the bottom of the haring aid. These two are the solutions that are commonly used.

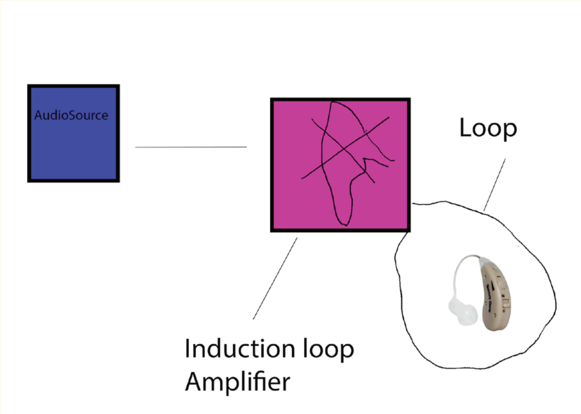


Image 1.33 showing the induction loop system

If the patient has near normal hearing in any frequency region, the wireless receiver should be coupled to the ear with an open earmold or ear shell so that the wireless system does not reduce unaided signal reception (namely sounds arriving acoustically from nearby talkers who are not wearing a transmitter. The open coupling will affect the gain frequency response of the complete system.

The digital signal processing in hearing aids can cause interference in a nearby wireless receiver which in turn can pass the interference on as an audio signal to the hearing aid which then produces interfering sounds in the user's ear. It is highly essential that a listening check be performed whenever a hearing aid and a wireless receiver are first coupled together.

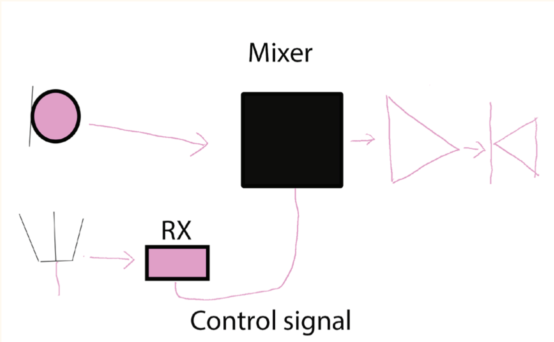


Image 1.34 showing a diagrammatic representation of a system in which the hearing aid automatically selects either just the local microphone signal or FM signal from the receiver (Rx) combined with an attenuated version of the local microphone signal.

Combining wireless and local microphones:

When it is necessary to hear more than one person talking, or need their hearing aids for own-voice monitoring, it is not satisfactory for the hearing aid to receive only the signal coming from the wireless transmitter, because the transmitter may be far away from the person speaking. Wireless systems overcome this problem by many ways. The common solution is for the wearer of the hearing aid to hear a mixture of sound coming from the transmitter and sound that is being picked up by the hearing aid microphone (the local microphone). While this allows a nearby speaker to be heard, the hearing aid microphone continues to pick up noise and reverberation even when the teacher is talking into the transmitter, thereby potentially removing most of the advantages provided by the wireless system. The problem is minimized by mixing two signals so that the signal from the FM system is more intense than the signal from the hearing aid microphone. This feature goes by the name FM advantage / FM priority.

There is a dilemma in setting up the mixture in a combined signal. The clearest signal from the teacher is received in the wireless alone condition, the worst signal from the teacher is obtained using the local microphone alone, and the signal of intermediate clarity is obtained in the combined position. The dilemma is that when the children are asked which operating mode they prefer, the order is reversed, presumably because the children feel increasingly detached from their environment as their local microphone becomes less dominant.

Ideal solution lies in an automatic switch within the wireless system that attenuates the hearing aid microphone when someone speaks into the transmitter microphone. It restores full sensitivity to the local microphone when the transmitter is not sending any signal. These systems are known as speech-operated switching (SOX) or voice operated switching (VOX).

The commonly used solution to this problems is a process known as dynamic FM. In this system the degree of FM advantage automatically increases as the background noise level as sensed by the transmitter increases.

The increased priority given to the FM signal increases speech intelligibility when there is background noise. This improvement in speech intelligibility in noise that is offered by wireless transmission is potentially very useful because the signal picked up by the transmitter microphone is likely to have SNR 20 dB greater than the SNR picked up by the hearing aid microphone.

Infra-red radiation

This is a type of electromagnetic energy similar to that of radio waves except for the fact that it occurs at a much higher frequency which is approximately 10^{14} Hz. Frequencies higher than this will be perceived by humans as a red light and hence the term infra-red. Audio signals can be transmitted by infra-red electromagnetic waves and it requires that the carrier wave (i.e which happens to be infrared wave) be modulated by the audio wave.

This system has proved to be useful in schools and Universities, in large auditoriums with partial signal coverage or in rooms where is a very high amount of radio-frequency emissions in varying frequencies.

Similar to FM signals the output of the infra-red system can directly drive a headphone or can easily be coupled electrically to a hearing aid. It is common for infra-red systems to be used directly with an earphone. Infra-red waves operates at the same frequencies as light waves. It hence travels in straight lines and can be blocked by opaque obstacles and would also reflect off a flat light colored surface. Direct sunlight is also known to interfere with transmission of infra-red waves. The receiver detects the light and demodulates the intensity fluctuations to recover information. If the recovered information is an audio wave then it can be presented to the headphones or coupled to a hearing aid.

Advantages

1. Transmission is reliable and absolutely free of interference caused by electromagnetic fields or structural elements in the buildings like metal reinforcements.

2. These systems are not known to produce electromagnetic emissions themselves

3. No registration and bandwidth allotment from Government authorities is needed as in the case of radio waves.

4. If the infra-red system is powerful enough then it can cover even a large room.

Classroom sound-field amplification

Sound field amplification systems delivers sound to the listener using acoustic waves that are propagated across the room. This technology works under the premise that if SPL, and hence SNR is adversely affected by distance from the talker and presence of background noises. These factors can be minimized by amplifying the wanted sound and positioning a loudspeaker near the listener. Common application for this technology is the classroom that contains a microphone, an amplifier and one or more loudspeakers. This is a very basic system. On obvious limitation is that the teacher needs to speak to the microphone which is either stationary (the teacher then needs to stand in front of it) or need to carry around the microphone that is attached to a long cable.

One essential addition to this technology is a radio-frequency like the FM, or infra-red link between the teacher and the amplifier which enables the teacher to move freely around the classroom.

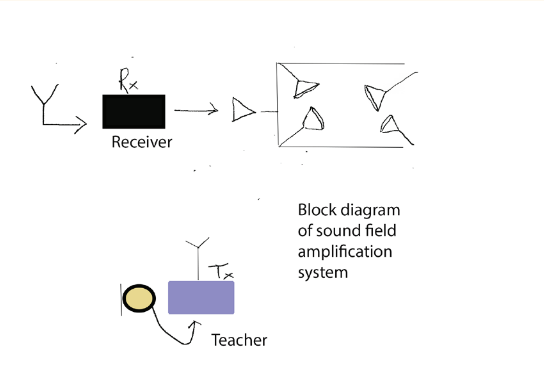


Image 1.35 showing a typical sound field amplification system

Advantages

1. This system does not require the listeners to wear any special equipment.

2. Installation is fairly simple and straight forward

3. System reliability is fairly high because nothing is worn on the body

4. Improved sound clarity is available for all children inside the room

5. Children with normal hearing will also benefit from a clear signal

6. Also helps children with conductive hearing loss due to wax or CSOM

This system is known to cause reverberations. There are of course ways to minimize this problem which include:

1. Loudspeakers can be positioned in the ceiling or high on the wall near each corner of the classroom so that as many children as possible are as close to one of the loudspeakers.

2. A directional vertical column loudspeaker at head height when seated can be used. These loudspeakers which contain an array to radiate much more sound horizontally than they do vertically.

3. When there are only a few children in the classroom who need assistance, each child needing assistance can have a loudspeaker positioned on their desk directly in front of them. This system is known as the desktop FM.

Sound field amplification is best for classroom setup because:

1. It increases the level of sound

2. It increases the ratio of direct to reverberant sound for anyone close to the loudspeaker than they are to the teacher.

3. It increases the signal to noise ratio SNR for everyone

4. Its benefits are immediate

5. It is immediately known to the teacher whenever there is a fault or glitch in the system

Assistive listening devices

These devices help hearing impaired persons detect sounds / understand speech, but are not worn on the head or body. These devices are used in conjunction with hearing aids or instead of hearing aids. The various wireless systems, radio systems and amplifier systems described previously come in this category.

Intelligibility is improved by locating a microphone closer to the source than is usually possible for

the user to be by directly connecting to the source device e.g. TV to a transmitter. Assisted learning devices that improve speech intelligibility can be classified according to their purpose:

One to one communication - These include sound field systems, infra-red systems and magnetic loop systems. In principle it is possible for the user to purchase a receiver and use it with transmitters available in public places like theaters, the sheer variety of system types like infra-red, radio frequency and various modulation types makes is unlikely that any single receiver would be compatible with multiple venues that the user visits frequently. Magnetic induction systems don't have this disadvantage, provided the hearing aid has a telecoil.

Television devices - These devices pick up the television audio via a plug and socket or use a microphone very close to the television speaker. The signal can be delivered to the user via a hard wired connection or a wireless link, which in turn can be either radio-frequency transmission or magnetic loop induction. The loop can be room-sized, chair-sized, or ear-sized.

Telephone devices:

These include -

An amplified telephone

An amplifier that is inserted between a regular telephone and its wall socket

A coupler that picks up the acoustic signal coming from the receiver and amplifies it to create a stronger acoustic output, a stronger magnetic output or an electrical signal that can drive a neck loop or a silhouette coil.

The increase in signal to noise ratio offered by these assisted listening devices with the microphone close to the source of sound is so huge that clients would benefit greatly if only these facilities are more frequently made available. The increased clarity that an assisted listening device can provide relative to acoustic reception across a noisy and reverberant room, will be useful for many patients.

ALD's (assisted listening devices) are also used to alert the user to environmental events. These are known as alerting ALD's and it comprises of a sensor of some type linked to any output that can easily be detected by the hearing impaired person. It could be in the form of a flashing light,vibrator or intense low frequency sound. The most common sensor triggers are:

A telephone ring sensor

A baby cry sensor

A smoke detector

An alarm clock

A door bell

The need of each patient at home, at work /school and for leisure should be reviewed. The need for each of these situations could be different. For example there may be needs for face to face communication, reception of media, telecommunications, and alerting to environmental sounds. These needs can easily be met with ALD's than by hearing aids alone.

Connectivity / convergence

People these days commonly spend significant amount of time wearing things in their ears. Most common of these devices include the earphones which enables them to listen to mobile phones and music players. There are also other electronic devices that produce audio signals. These devices include radio receivers, home entertainment systems, portable video players, personal digital assistants and computers. Persons with hearing loss have the same needs as persons with normal hearing and they need to listen and communicate. It is really inconvenient to remove a hearing aid and insert an earphone every now and then. Removal of the hearing aid also takes away the individually shaped gain-frequency response to the detriment of intelligibility ad tonal quality especially for persons with sensorineural hearing loss. Hence there is a growing need for hearing aids to connect easily to other devices or to fulfill the function of these other devices.

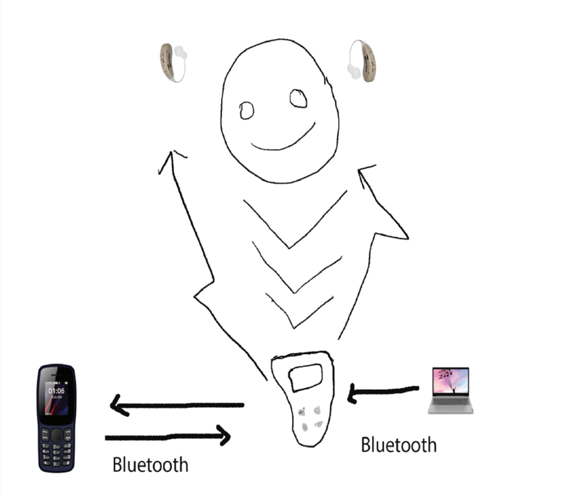


Image 1.36 showing connectivity between hearing aids and various peripheral devices

Interference between mobile phones and hearing aids

Interference of the hearing aid by mobile phones has been a major problem during the past decade. But with advances in technology this is slowly decreasing. Interference is not surprising as mobile phones are designed to transmit strong electromagnetic signals and hearing aids contain many conductors that could act as aerials when immersed in an electromagnetic field. The resistance of hearing aids to interference has progressively reduced so at most interference is caused when the user holds the phone close to the head.

The level of interference depends on:

1. Design of the hearing aid

2. Carrier frequency and modulation method used in the telephone transmission system

3. The distance of the mobile phone from the nearest phone tower and the presence of intervening obstacles

4. The design of the phone (both location of the aerial and layout of conductors carrying current)

5. The distance between the telephone and the hearing aid

6. Orientation of each device in relation to the other

Patients hence should be advised to try a mobile phone with their hearing aid before finalizing the purchase. Ideally the trial should be conducted in a region of low tower strength so that the mobile

is transmitting at maximum power. The lower the IRIL (input related interference level) value for the hearing aid better it would suit the hearing aid user.

Hearing aid Performance

Performance of hearing aids can conveniently be measured when it is connected to a coupler. A coupler is a small cavity that connects the hearing aid sound outlet to a measurement microphone. The standard 2 cc coupler available is larger than the average adult ear canal with hearing aid in place. This difference is termed as the real-ear-to-coupler difference (RECD). This quantity is worth measuring in infants because they have ear canals that are different from that of an average adult.

Test boxes ensure that sound reaches the ear in a controlled manner for the purpose of testing. These sounds can be pure tones that sweep in frequency or can be complex broadband sound like speech which could contain many frequencies simultaneously. Broadband sounds are necessary to perform meaningful measurements on many nonlinear hearing aids. The test sound should also approximate the spectral and temporal properties of speech so that the various signal processing algorithms in the hearing aid alter the gain in a manner representative of actual use.

The measurements commonly performed using test sounds are curves of gain or output versus frequency at different input frequencies. The curve of output versus frequency when measured with a 90 dB SPL pure tone input level is taken to represent the highest levels that a hearing aid can create. Other test box measurements that are less commonly performed are measures of distortion, internal noise, and response to magnetic fields. These measurements are used to check to ensure that hearing aid is operating according to its specifications.

Test box measurements are but a means to an end. The end is the performance of the hearing aid in an individual's ear. This performance can be directly measured by using a soft, thin probe-tube inserted into the ear canal. Real-ear performance can be expressed as real-ear aided response.

Real-ear-aided gain (REAG) is the level of sound in the ear canal minus the input level of sound near the patient or as real-ear insertion gain (REIG) the level of sound in the ear canal when aided minus the level in the same place when no hearing aid is worn. Each of these values requires the probe to be carefully located.

Both these types of real-ear gain are different from that of coupler gain because of the difference in the volume of the ear canal and that of the coupler. Moreover the input to the hearing aid microphone is affected by sound diffraction patterns around the head and the ear. The changes in SPL (sound pressure level) caused by diffraction are referred to as microphone location effects.

Factors that lead to incorrect measurement of real-ear gain

Incorrect positioning of the probe

Squashing of the probe

Blockage of probe by cerumen

Background noise

Saturation of hearing aid

Measuring hearing aids in couplers and ear simulators

Hearing aids can be conveniently measured in couplers and ear simulators. Availability of standard couplers and simulators allows measurements to be made in different places and at different times under identical conditions.

Couplers and simulators

A coupler is a cavity. It has the hearing aid connected to one end and a microphone connected to the other. This equipment provides a repeatable way to connect the hearing aid to a microphone, and hence to a sound level meter without sounds leaking out to other places. The standard coupler commonly used for hearing aids has been in use for more than 50 years and has a standard volume of 2cc.

This volume is chosen because it approximated the volume of adult ear canal past the earmold when a hearing aid is worn (i.e. the residual ear canal volume) and the equivalent volume of the ear drum and middle ear. It is not an accurate approximation of the acoustic impedance of the ear at high frequencies.

The sound pressure level (SPL) generated in any cavity by a hearing aid depends directly on the impedance of the cavity, this value in turn depends on the volume of the cavity. In an average adult ear, the residual ear canal has a physical volume of about 0.5 cc. This volume acts as an acoustic spring, or acoustic compliance. The ear canal terminates in the ear drum, on the other side of which lies the middle ear cavity. The compliance value of middle ear cavity and ear drum together act as if they have a volume of 0.8 cc. The combined 1.3 cc volume determines the impedance for low frequency sounds. As the frequency increases, the mass of the eardrum and ossicles causes their impedance to increase, while the impedance of the residual ear canal volume falls. Hence for increasing frequency, the total impedance does not decrease as much as would be expected for a simple cavity.

An ear simulator mimics the ear's variation of impedance with frequency. Image 1.37 shows the concept behind one ear simulator. This simulator has a main cavity with a volume of 0.6 cc. This simulator has four cavites present at its side each one measuring in volume between 0.1 - 0.22 cc. These cavities are connected to the main cavity by small tubes. Three of these tubes contain dampers. As the frequency increases, the impedance of these tubes rise and they close off, thereby causing the effective total volume to gradually fall from 1.3 cc to 0.6 cc.

Zwislocki coupler is a similar coupler that is available commercially. Other ear simulators like Bruel and Kjaer operate on similar principles, except for the fact that they have two side cavities instead of four as is seen in Zwislocki coupler.

Several standards have been published by the American national standards institute (ANSI) and International electro-technical commission (IEC) specify how hearing aids should be tested. These various standards allow 2-cc couplers to be used as ear simulators. In order to understand and interpret clearing a hearing aid specification sheet, it is essential to determine whether thee data refers to coupler or ear simulator performance because the gain of the coupler could be higher than that of ear simulator, and also whether the hearing aid has been measured in a test box or on an acoustic manikin. An acoustic manikin consists of a head and torso with an ear simulator incorporated inside each ear. The choices of coupler versus ear simulator, and test box versus manikin make a big difference to the numbers specified.

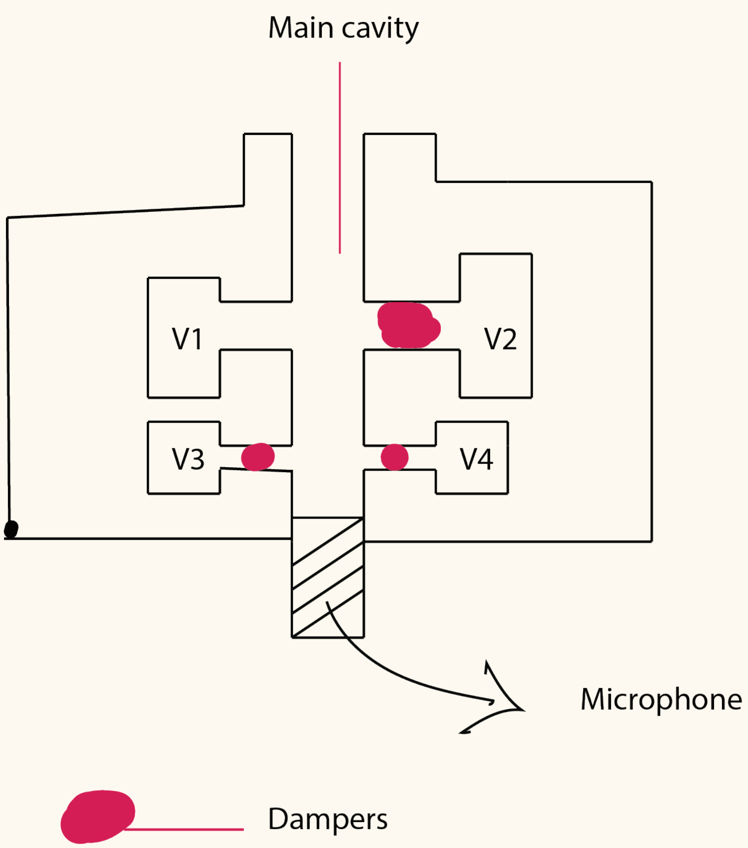


Image 1.37 showing a simplified internal structure of a four-branch ear stimulator

Couplers and ear simulators will have to be connected to any type of hearing aid for performing measurements. In order to achieve this, a range of adapters can be used.

Reference plane:

This is a plane which is at right angles to the longitudinal axis of the ear canal, located at the point in the ear canal where the earmold or ear shell usually terminates. This is approximately 13 mm from the ear drum. Hearing aids usually terminates at this point. BTE hearing aids use tubing to connect to the external canal, so they also require tubing when connecting to a coupler or simulator.

Types of couplers as per ANSI S 3.7 standards:

HA1 coupler:

This coupler has no earmold simulator and is used for ITE and ITC aids, which are connected to the coupler via putty. It can also be used for BTE hearing aids not intended to be used with ear-molds (thin tube BTE's that terminate in a dome) or RITE BTE's.

HA2 coupler:

Includes an earmold simulator which is connected to the BTE hearing aid via tubing or into which a receiver for a body aid snaps. The H4A coupler is a variation of the HA2 coupler intended for BTE or spectacle aids in which the tubing diameter from the hearing aid to the medial tip of the earmold is a constant 2 mm diameter. Use of HA4 coupler is less common these days.

Ear simulator is better than 2-cc coupler in that it has the same variation of impedance with frequency as the average ear, a hearing aid generates the same SPL in an ear simulator as it does when inserted to the reference plane in an average adult ear. This equivalency assumes that the hearing aid is coupled to the ear simulator in the same way that it is coupled to the ear canal. It should be pointed out that even ear simulator is not accurate enough to measure the sound pressure level that would be present in an individual ear.

Connection methods for ear simulators are similar to those used for 2-cc couplers. The connection option for the ear simulator is an ear canal extension attached to the opening of the ear simulator at the reference plane. The dome of a thin tube instant fit BTE hearing aid is inserted into the canal simulator just as it would be inserted into the ear.

Disadvantages of ear simulator compared to 2-cc coupler include:

1. Cost is high

2. Potential for the small openings inside the simulator to become blocked

Ear couplers and simulators will not produce accurate results if:

1. The sound bore of an hearing aid is poorly sealed to the coupler or simulator

2. The tubing connecting to a BTE hearing aid becomes stiff and does not properly seal at either end

3. The O-ring that connects a button receiver wears out

4. Pressure equalization hold becomes blocked or is excessively open.

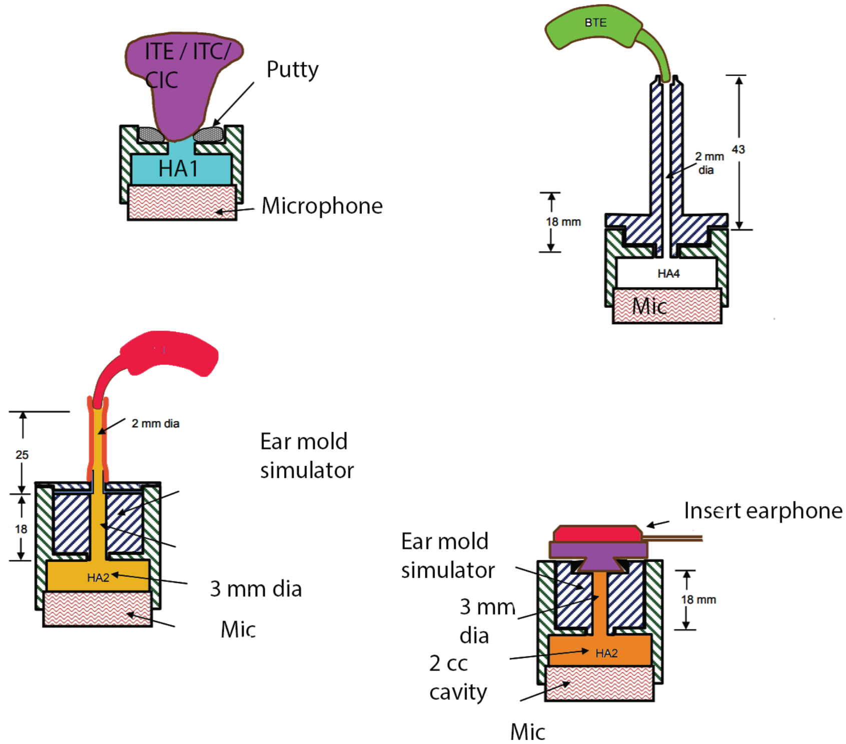


Image 1.38 showing various types of couplers used in measuring SPL in hearing aids

All the above stated defects except for that of the blocked equalization hole, will all decrease the low frequency gain. Measurements taken will create a spurious mid-frequency resonance.

Bone conduction hearing aid can be measured using artificial mastoid. It provides a way of measuring the force output of a bone conductor hearing aid over the frequency range of 125 to 8000 Hz.
Test box:

Generates sounds of a required SPL at the hearing aid microphone. Components of a test box include a tone/noise generator, an amplifier, a loudspeaker and a control microphone. The control microphone is also known as the reference microphone is placed next to the hearing aid microphone. This control microphone monitors the sound pressure level (SPL) reaching the hearing aid from the loudspeaker. If the input level is higher or lower than the desired level, the control microphone circuit automatically turns the volume of the sound

coming from the test box speaker either up or down until the required level is obtained.

Control microphone works in either of the two ways:

Pressure method - The control microphone is placed as close as possible to the hearing aid microphone while the measurement is being performed. The control microphone corrects the field during every measurement.

Substitution method - The control microphone is placed in the test position prior to the actual measurement. During a calibration measurement, the control microphone measures the SPL present at each frequency, and stores any discrepancy between the actual and desired SPLs. During all subsequent measurements, the test box adjusts its outputs to compensate for these discrepancies.

The test box also attenuates ambient noise by having a lid that seals well to the box. As it is constructed with solid, dense walls and is also filled with absorbent material inside. It also minimizes the amount of reflected sound that reaches the hearing aid. Test boxes measures either of the two different types of signals. The traditional measurement signal is a pure tone that automatically sweeps in frequency over the desired frequency range (between 125 Hz - 8 kHz). It is more appropriate to measure modern hearing aids with broadband test signals. These signals have a wide range of frequencies and are presented simultaneously. The test box uses a Fourier Transform or a swept filter to determine the level of each frequency component coming out of the hearing aid. Since the analyzer stores the level of each frequency component at the input to the hearing aid, it can calculate the gain at each frequency. If the hearing aid is operating linearly, measurement with pure tones will give exactly the same gain-frequency response as measurement with a broadband signal.

Majority of hearing aids don't amplify linearly over a wide range of input levels. The most common cause of nonlinearity is compression which involves an amplifier whose gain depends on the input signal.

Realistic assessment of the effect of hearing aid on speech will occur when the input spectrum has a spectrum similar to that of speech frequency. Broadband signals used in test boxes usually have such a spectrum.

Test signals used include:

1. Spectrally shaped random noise

2. A repetitive waveform with a crest factor which is similar to that of speech

3. A series of short tone bursts that vary rapidly in frequency and amplitude to match both the spectrum and dynamic range of speech

4. Speech sounds that have been processed to remove the fine detail that provides most of the intelligibility while retaining the temporal fluctuations in amplitude of real speech such as on for the ICRA noises

5. Speech syllables extracted from multiple languages and pasted together to sound like speech, this is referred to as the International speech test signal (ISTS).

6. Actual continuous speech

For hearing aids with multiple bands of compression, gain measured with pure tone sweep will be less than the gain measured with a speech-shaped signal. The more advanced the hearing aid being tested, it is more important for the test signal to simulate those characteristics of real speech used by the hearing aid to control its amplification characteristics.

Measurements commonly performed on hearing aids are gain frequency response and OSPL 90-frequency response. Image 1.39 shows an example of each of these responses obtained with a BTE model hearing aid in an HA2 style- 2-cc coupler and measured with a swept pure tone signal. The gain frequency response was obtained with an input signal level of 60 dB SPL. The results are plotted in vertical axes. The left hand axis shows the output in dB SPL. The gain at any frequency can be calculated as the output SPL at that frequency minus the input SPL. This gain is directly shown on the right hand axis in the figure.

Terms that are used to summarize the gain-frequency response:

High frequency average gain (HFA gain): Average of gains at 1000, 1600 and 2500 Hz.

Special purpose average gain: Average of the gains at three frequencies, each separated by 2/3 octaves. This is used for hearing aid with unusual frequency responses.

Frequency range: This is the range of frequencies between the lowest and highest frequencies whose gains are 20 dB below the HFA gain.

IEC and ANSI standards specify that the hearing aid maximum output should be measured using a 90 dB SPL input signal and both these standards use the term OSPL90 to describe this measurement. This level is considered to be high enough to cause most hearing aids to reach their highest possible output level at each frequency. When the hearing aid output has reached its limit for any

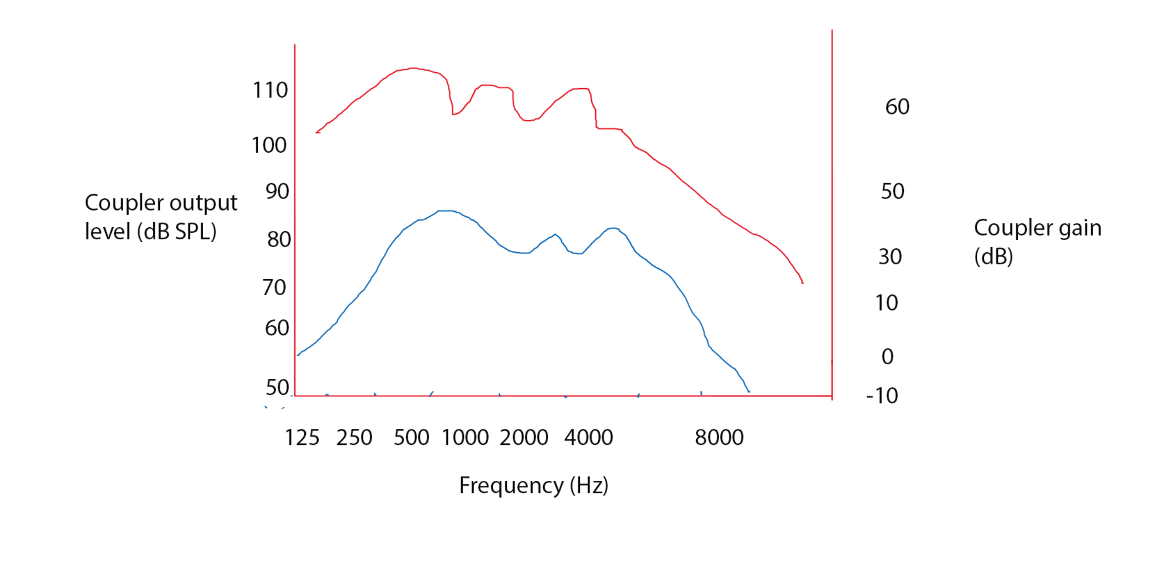


Image 1.39 showing the gain frequency repose measured with a 60 dB SPL input level

input signal, it is said to be saturated. Hearing aids with a steeply rising response will often not be saturated at low frequencies.

The amount of gain measured with a hearing aid depends on where the volume control and all the filter controls are set. If the volume control is in the full-on position, a full-on gain is obtained, or else it should be set at the reference-test setting in which case the resulting gain curve is referred to as the basic frequency response. The purpose of reducing the volume control to a reference position is to set the hearing aid so that it is not saturated for mid-level input signals.

Input output functions

Input output function shows the output level versus input level for one frequency or for one broadband test signal. All hearing aids become nonlinear at high input levels, and many are non-linear over a wide range of input levels the I-O function is a valuable tool for understanding how a hearing aid modifies sound.

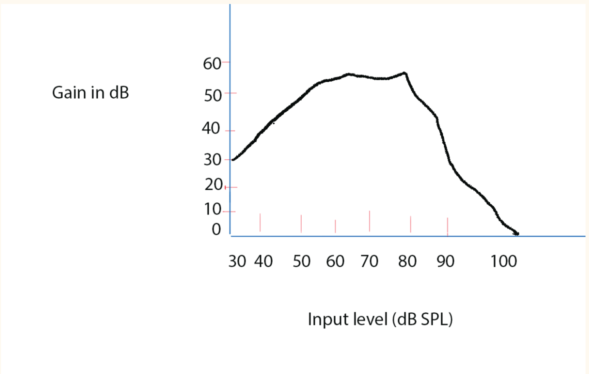


Image 1.40 showing gain output diagram

Gain:

This is the amount of sound the hearing aid adds. It is the difference between the input level and output in dB SPL. It is typical for the gain of linear hearing aids to be measured at 60 dB and also at the maximum volume wheel setting (full gain setting). for non-linear hearing aids, it is customary to measure the gain at low, medium and high input levels. These values are arrived at by comparison with the input signal.

Compression & Expansion:

They describe the effect of the amplifier on the dynamic range of the signal that varies in amplitude over time. A compressor makes the signal's dynamic range smaller, regardless of whether the output level is smaller or larger than the input level. In contrast the expander increases the dynamic range.

A hearing aid can simultaneously:

Amplify and compress

Amplify and expand

Attenuate and compress

Attenuate and expand

The input sound. Magnetic response can be measured easily if the test box contains a loop to generate a magnetic field. If it does not contain a loop then it is impossible to measure. The precautions taken while measuring magnetic response include:

1. The volume control should be at its reference position when measuring magnetic frequency response.

2. The hearing aid should be oriented as it would normally be used regularly.

Since telephones happens to be an important source of magnetic signals ANSI specifies standards for telephone magnetic field simulator that generates magnetic signals similar in level and field shape pattern to those generated by telephone.

Real-Ear-to-coupler-Difference:

The difference between the SPL a hearing aid delivers to the ear canal and the SPL it delivers to a coupler is called the real-ear-to-coupler difference RECD.

RECD = Canal SPL - Coupler SPL

RECD concept is useful in hearing aid fitting and audiometry, that too for fitting hearing aids to babies whose ear canals are small. Accurate knowledge of this value allows the audiologist to interpret hearing thresholds measured with insert earphones, and allows for a more accurate adjustment of hearing aid in a coupler so that it achieves the desired performance in the patient's ear.

Factors affecting RECD:

Ear canal volume - The ear canal volume is smaller than that of the standard coupler which measures about 2 cc. Female ears are smaller than that of males and hence female RECD values are about 1-2 dB higher than male RECDs. The difference between real ears and a 2-cc coupler occurs even for a standard insertion depth in the real ear. If the earmold is inserted deeper into the ear canal, the residual volume could get even smaller, and the SPL generated will even be greater and hence the RECD will be larger for these occluding earmolds.

Leakage, vents and open fittings - Earmolds very rarely fit in the ear without leakage. The leakage can also be intentional or may be accidental. On the other hand it is easy to connect a tube to a coupler with zero leakage. Leaks and vents allow low frequency sound out of the ear canal causing a smaller SPL within the ear canal and thus reducing he RECD in the low frequencies relative to a well sealed ear canal. Leakage can be so great that RECD for 250 Hz could become negative. There is a more likelihood of greater leak in custom molds than foam ear tips that expands within the ear canal.

Open canal fitting of hearing aid can be considered as an extremely large vent and hence the RECD value could be negative.

Tubing - The diameter of tubing affects the amount of vibrating air that flows back and forth between the tubing and the ear canal or coupler. This flow is technically known as the volume velocity.

Transducer type - RECD value will reflect the differing volume of the ear canal relative to the coupler only if the transducer and tubing system used delivers the same volume velocity in each case. Transducers used in insert earphone that is often used to measure RECD does not have an impedance many times higher than that of ear canals and couplers at all frequencies. BTE hearing aids can also be used as the transducer, but if they h ave no damping in th ear hook, they may even have an impedance lower than that of the ear around their resonant frequencies. RECD values can vary up to 10 dB across transducers.

Usefulness of RECD values:

RECD is likely to be useful if the ear canal is significantly different from that of an average adult. Two categories of patients for whom differences are more likely are children under the age of 5 and people

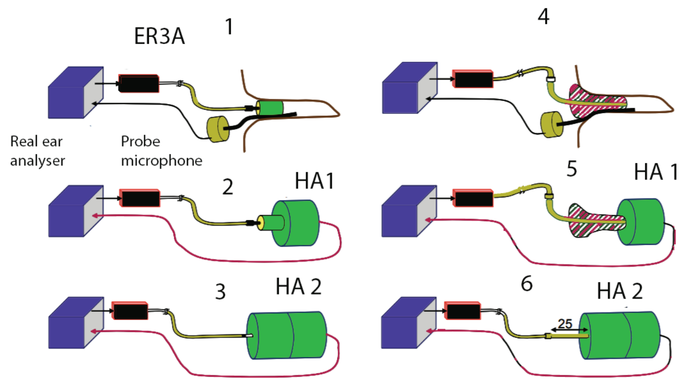


Image 1.41 showing measurement of RECD using real ear analyzer. The insert phone is connected to the patient's ear 1. via foam tip 2. HA1 coupler via a foam tip 3. Directly to a HA2 coupler. For BTE's that will be fit with a custom earmold the insert phone is connected to 4. Patient's ear by individual earmold and tubing and 5. a HA1 coupler via individual earmold and tubing and 6. A HA2 coupler by 25 mm of tubing. The RECD values obtained with a HA1 coupler will be different from those obtained with a HA2 coupler.

who have had surgery on their ear canals. RECD can easily be measured using real ear gain analyzer. The microphone attached to the probe records the SPL when attached to the external auditory canal.

If RECD is measured for BTE hearing aid then in all probability the patient will be using a custom earmold. So it is better to include the patient's actual earmold in the measurement.

Alternate method of measuring RECD

Some hearing aids have features for measuring RECD in conjunction with the software use to program the hearing aid. The measurement can be enabled by positioning the probe tube in the ear canal with the other end of the probe tube connected to:

1. A hearing aid microphone inside the faceplate of an ITE hearing aid or case of a BTE hearing aid

2. A special boot plugged into the direct audio input of the BTE hearing aid.

Since patients commonly have similar earmolds in each ear because of symmetrical ear canals, RECD values for the two ears usually match and is within 3 dB for both children and adults. If time does not permit or if the child is uncooperative then measurement from one ear can be used to the other ear hearing aid also.

Causes of errors in RECD measurements:

1. High frequency gain- This error occurs if the probe is not inserted sufficiently deeply or if the wrong type of 2-cc coupler is used

2. Mid-frequency gain - This error occurs if an inappropriate transducer is used for the measurement

3. Low frequency gain - This error occurs if sound leaks through a vent, and sound directly entering the ear canal via a vent are not properly factorized for.

Real ear to Dial Difference (REDD)

This quantity is equal to the SPL in the ear minus the setting (in dB HL) on the dial of an audiometer.

REDD = Canal SPL - Dial HL

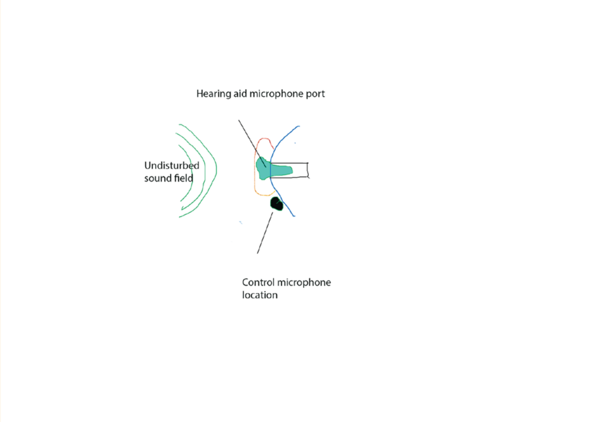
In a properly calibrated audiometer, the SPL in the coupler used to calibrate the earphones will be equal to the dial setting plus the Reference Equivalent Threshold SPL RETSPL. RETSPL are known for commonly used headphones, insert earphones and couplers.

Coupler SPL = Dial HL + RETSPL

Real ear aided gain (REAG):

It is the quantum of gain of the hearing aid in the individual's ear that is important. Real ear gain can be estimated from coupler gain, but there is always some degree of inaccuracy in the estimate caused by variations in ear sizes, fit of the hearing aid or mold to the ear canal, size of the sound tube and vent path and the location of the microphone relative to the pinna. There is also some amount of small variation between identical hearing aids. Measurement of individual real ear gain is important unless the hearing aid fitting software can predict it within about 5 dB.

This value can be measured by placing a probe tube, connected to a microphone into the ear canal. There are two different types of real-ear gain. The first happens to be the real ear aided gain and the next is real ear insertion gain. The real ear aided gain which is expressed in dB, is defined as the SPL near the eardrum, minus the SPL at some reference point outside the head. This reference point is defined as the level in the undisturbed field, F or the level at a control microphone mounted on the surface of the head as shown in the figure below.



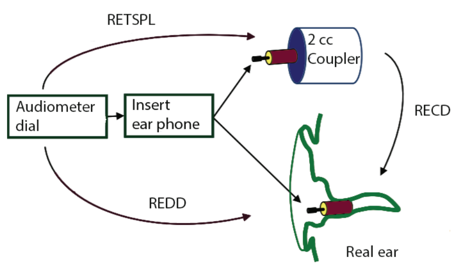


Image 1.42 showing the relationship between RETSPL (coupler minus dial), RECD (real ear minus coupler), and REDD (real ear minus dial).

The real ear measurement equipment uses the signal from the control microphone to regulate the sound level near the ear to the required level. The control microphone removes the diffraction effects from the free field to the surface of the head. These diffraction effects are minimal when the sound source is located directly in-front of the aid wearer.

Synonyms for real ear aided gain include:

Real ear aided response

In-situ gain

Real ear transmission gain

Probe positioning for REAG measurement

This depends on the type of equipment used. In all cases a flexible probe is inserted into the ear canal so that the SPL in the residual canal is sensed while the hearing aid is in place and operating. The probe is inserted first, and then the hearing aid / earmold. Most important part is to obtain the correct depth of insertion.

A marker is placed on the probe tube about 30 mm from the open end. A continuous tone is generated at 6 kHz and the probe is moved inwards smoothly and continuously starting at the entrance of the ear canal while the SPL is sensed by the probe microphone. The position at which the SPL is a minimum is identified. This location would be 15 mm from the acoustic Centre of the ear drum. This is the location at which the probe should be placed.

Insertion gain

This is the second type of real-ear gain. It is also known as real-ear insertion gain (REIG). This gain estimates how much extra sound is presented to the ear drum as a result of inserting the hearing aid in the ear. The main distinction between insertion gain and REAG is that insertion gain takes into account the amount of amplification the person is getting from the resonances in his / her concha and ear canal prior to insertion of hearing aid. This natural amplification is known as the real-ear unaided gain (REUG). This is lost either partially or wholly depending on how open the earmold is when the hearing aid is inserted. Before a hearing aid can provide additional signal, it must first provide at least this much gain.

Positioning probe for insertion gain measurement

The positioning of the probe for insertion gain measurement is not that critical as is the case for REAG measurement. Even though the interest is on the increase in SPL at the ear drum caused by inserting the hearing aid, the same increase will also occur at other points within the ear canal medial to the tip of the mod or aid. The SPL increase at the mid-canal position doesn't depend on the source of the sound.

1. The ear canal is inspected for presence of wax and other abnormalities

2. Aid / mold is inserted in the ear. The location where its lateral surface lies with respect to some landmark on the ear is noted. The ear canal entrance for inside the ear hearing aid, or the inter-tragal notch or tragus for larger aids.

3. The aid / mold is removed and the probe is inserted along the inferior surface of the external canal. The tip of the probe should be placed in such a way that it would extend 5 mm past the tip of the hearing aid / mold to avoid transition sound field.

4. This position is marked on the probe using a sliding marker

5. The probe is inserted until the marker lines up with the selected landmark on the ear and REUG value is measured

6. Now the hearing aid is inserted leaving the probe tip in the same position and REAG is measured.

Insertion gain will differ from that of coupler gain. It exceeds that of the coupler gain because it h as the benefit of head, pinna, and concha diffraction. Added to this factor the volume of the residual ear canal is smaller than that of a 2-cc coupler. Since this measurement is performed in two stages (unaided and aided) this would help in spotting errors at an early stage.

The difference between coupler gain and insertion gain, measured with no venting and with the same sound bore in each measurement is referred to as CORFIG (Coupler response for flat insertion gain). CORFIG values relate insertion gain to coupler gain at the same position of the volume control. This value is often used to find the coupler gain that is equivalent to a certain insertion gain. These coupler gains are used to select an appropriate hearing aid / to adjust the hearing aid during hearing aid trial.

In the case of In the canal hearing aids, up to 3 kHz, insertion gain approximately equals coupler gain for an average adult. For other types of hearing aids, there is a net difference between the insertion gain and coupler gain even on an average.

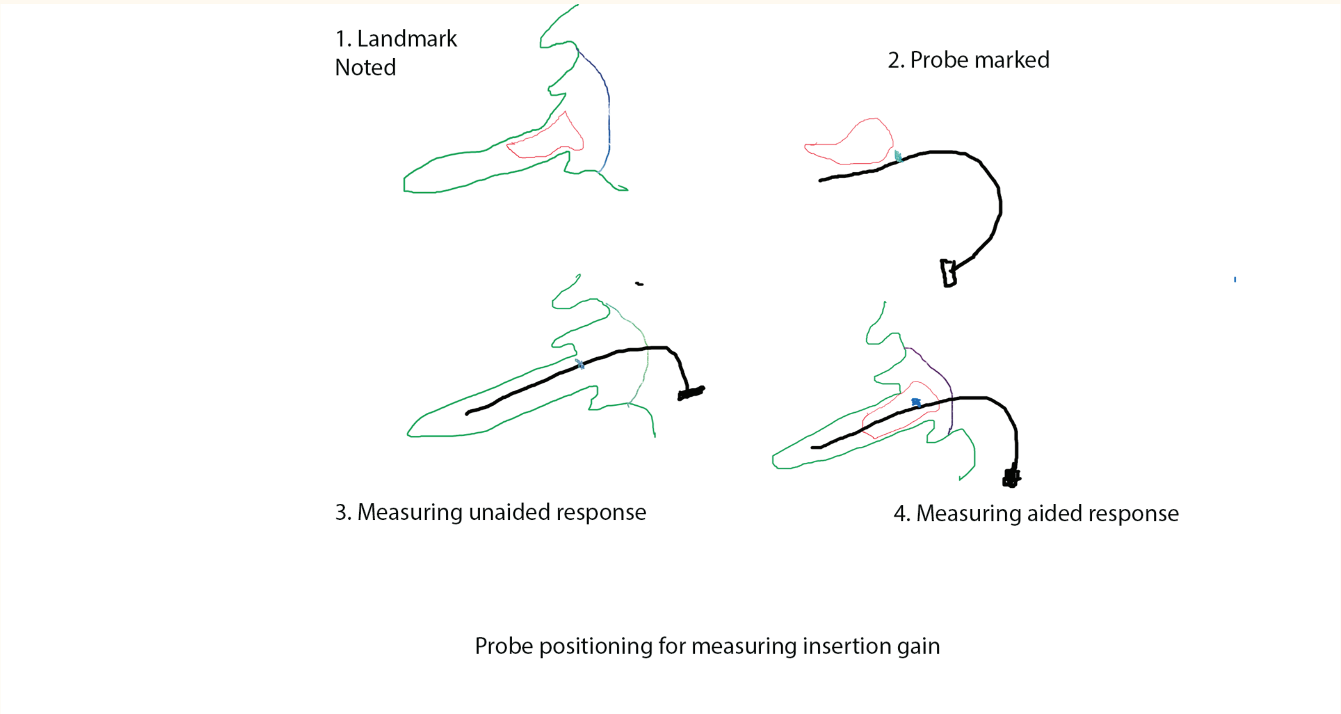


Image 1.43 showing the process of measuring insertion gain

Incorrect insertion gain measurements can easily be identified if the features given below are scrupulously looked out for:

1. There must be a low frequency plateau at the level of the test stimulus.

2. There must be a peak between 2.2 kHz and 3.2 kHz with an elevation above the low frequency plateau between 12 and 22 dB.

Accuracy of insertion gain measurements

How accurate are insertion gain measurements?

This can be arrived at by performing repeated measurements made with a variety of measurement methods. Over a majority of frequency range, the difference between a single measurement and the average of many measurements has a standard deviation of 3 dB. This means that 95% of measurements would be within 6 dB of the true value. In high frequencies because of the effects of standing waves and the impossibility of placing the probe exactly in the same place for the aided and unaided measurements, the standard deviation increases to 5 dB.

Issues that should be considered while performing insertion gain measurement.

Probe calibration:

Probe microphones have a non-flat frequency response because of the long thin probe tube. Real-ear gain equipment corrects for the response by including a calibration step in the measurement or by applying a correction step stored in memory. The probe microphone is calibrated against the control microphone which has a flat response

curve. In this procedure the clinician should hold the tip of the probe tune closely against the control microphone inlet port, but without blocking the inlet of either microphone. If the measurement system does not have a special clip to hold the two microphones together, it can be done with putty or with the fingers.

Control microphones:

This is also known as the reference microphone and is used to regulate the input SPL to the desired value. Commonly the pressure method of calibration is used in which the control microphone operates while the actual measurement is being made. If the hearing aid wearer moves between aided and unaided parts of the measurement, the control microphone compensates for the movement thus avoiding the measurement error that could occur.

The signal sensed by the control microphone will be greater than that amplified by the hearing aid, causing an artificially low gain measurement. This error increases with the gain of the hearing aid. This can be overcome by turning off the control microphone during the aided measurement.

Effects of wax:

Wax can cause errors during real ear gain measurements when it fills the tip of the probe tube. The equipment the incorrectly indicates that the signal level in the ear is pretty low. Cerumen should not have much effect on low-frequency real ear gain until there is enough of it to fill a significant portion of the residual ear canal volume. It should not have much effect on high frequency real ear gain until there is enough of it to fill a significant proportion (nearly one third) of the cross sectional area of the canal at any point.

Contamination by background noise:

Real ear gain measurement equipment employs a filter to dampen the background noise. For a broad band stimuli, the process of analysis follows a Fourier transform. Some of the testing equipment employs signal averaging to improve accuracy of measurement. Because of these factors real ear gain measurement system are resistant to ambient noise.

The tester should identify the lowest signal level at which measurement is possible and then avoid testing at lower levels. The ability to test at 65 dB SPL is the bare minimum that is acceptable.

Aided threshold testing and Functional Gain

Before the introduction of probe-tube equipment, hearing aid real ear gain was tested by finding the hearing thresholds in a sound field while the person was using the hearing aid and when not using it. The difference between these two values is known as the functional gain. This value more or less represents the insertion gain. If the hearing aid is operating in a nonlinear region for measurement then insertion gain and functional gain are equal only if the insertion gain is measured with the input level equal to the aided threshold.

For insertion gain, the field level is the same for the unaided and aided measurements, and the acoustic effect of inserting the hearing aid on the eardrum SPL is measured.

For functional gain, the ear drum level is the same for the unaided and aided measurements and the acoustic effect of inserting the hearing aid on field SPL is measured.

Advantages of insertion gain over functional gain measurements:

1. Insertion gain is more accurate

2. It can be measured in less time

3. Gives results at many finely spaced frequencies instead of just the audiometric frequencies

4. Can be measured at a range of input levels

5. It is not affected by the problem of masked aided thresholds

6. It requires the hearing aid wearer to sit still

One big disadvantage of aided threshold testing is that for people with near normal hearing at any frequency, aided thresholds will often be invalid.

Feedback in Hearing aids

This is a whistling sound heard by the user when the hearing aid is inserted. The term feedback means that some of the output of the hearing aid manages to get back to the input of the aid. When it reaches the input, it is amplified along with other signals that are received. This feedback impulses gets stronger and stronger as it goes through the loop of the hearing aid. This process would stop only when the signal is so strong that the hearing aid changes its operating characteristics sufficiently because the signal has grown too large.

In the case of linear hearing aids, this would be when the output limits the peak by peak clipping or by compression limiting. In a nonlinear hearing aid, it may be when the gain of the hearing aid decreases because of compression. Until this process of limiting occurs, the signal grows everytime it passes around the loop no matter how small the original signal was. It should also be pointed out that for feedback to be generated there is no need for an original signal to exist, just a small random sound is sufficient to start this process. These sounds are always present in the environment.

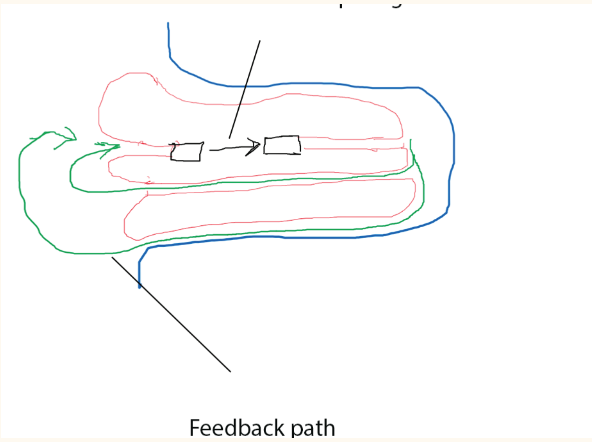


Image 1.44 showing feedback loop in a hearing aid

Feedback occurs only if the amount of amplification through the hearing aid is grater than the amount of attenuation from the ear canal back to the microphone. So only if the real ear aided gain of the hearing aid is greater than the attenuation can continuous feedback oscillations would occur.

Common causes of hearing aid feedback:

1. Positioning of a sound reflector near the hearing aid (a telephone)

2. Talking / chewing causing ear canal changes including changes in its shape could create a sound path past the earmold increasing the chances of

feedback

3. Growth of the ear canal (common in children) can increase the amount of sound leak between the earmold and the ear canal

4. Shrinkage of earmold when it becomes old.

Effect of feedback on sound quality

Excessive feedback has two important adverse effects. The first and foremost is thee audible whistling which is the most obvious, and it could be heard by everyone inside the room except the person wearing the aid. This could happen if the aid wearer has a hearing loss at a frequency that even maximum output from the hearing aid at this frequency is inaudible. Currently the widespread use of automatic feedback cancellation circuits have made this scenario a thing of he past. A hearing aid can oscillate at a frequency only if there is enough gain at that frequency and there is no point in providing gain if the aid wearer cannot hear a signal at maximum output SPL at that frequency. The current crop of hearing aids are much more flexible making it easy to decrease the gain in specific frequency regions to avoid this problem.

The second problem is a little subtle. When the hearing aid gain is set a few dB below the point at which the aid continually oscillates, the signal feeding back will still cause the gain to increase at frequencies where the feedback is positive.

Feedback induces extra peakiness in the hearing aid response and these peaks occur at the potential feedback frequencies. Every time a sound with components at these frequencies is put into the hearing aid, the hearing aid rings for a little while after the signal has ceased. The ring mechanism is very similar to the reason why a bell continues to vibrate and ring after it has been struck. The bell (in this case the hearing aid) stores energy and gradually releases it at this frequency over the next few hundredths of a second). This ringing effect is known as sub-oscillatory feedback.

The increased peakiness and the ringing effect rapidly decreases as the gain of the hearing aid is decreased below the point at which feedback oscillation becomes continuous. A 5-6 dB of gain reduction could be sufficient to reduce this effect.

Probe tube inserted for real ear gain measurements can cause feedback. Inserting the probe between the mold and the canal wall creates small additional leakage paths on either side of the probe. A hearing aid hence could whistle when it is being measured but could be totally satisfactory otherwise. Even if there is no leakage around the probe tube, there can be leakage through the wall of the probe. The tip of the probe is in the residual ear canal and so the full output of the hearing aid exists at all points within the probe tube. This high level acoustical signal vibrates the walls of the prob and hence the air outside the probe near the hearing aid microphone. Both these leakage paths are considered to be significant only for high gain hearing aids and in some CIC aids.

All other types of aids would have molds sufficiently loose and the extra leakage created around the tube is insignificant.

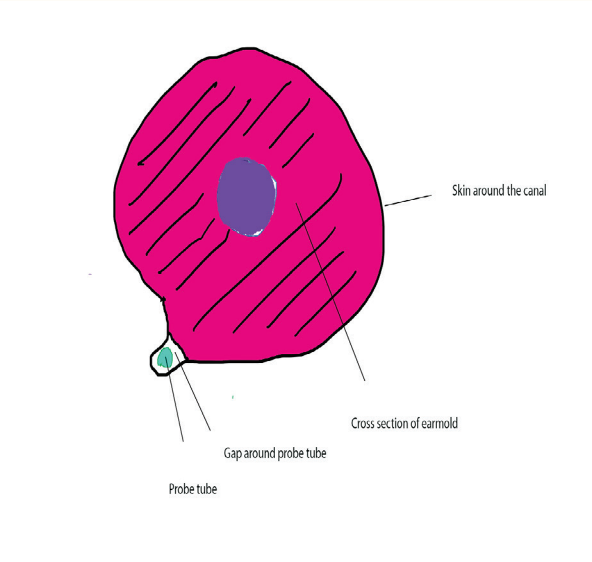


Image 1.45 showing leak around probe placed in the external canal

Troubleshooting faulty hearing aids

Even though this is not a domain of the clinician they are expected to try out simple troubleshooting tips before sending the hearing aid to the company for repair. They are not expected to carry out major repairs, many faults in the hearing aid happens to be trivial so simple steps can be followed to rectify them. When the patient complains of a faulty hearing aid, it is the responsibility of the clinician to decide whether to return the hearing aid to the company for repairs. It is of course unnecessarily troublesome to return the hearing aid to the company when repair could be been done on the spot within a matter of minutes.

As a first step in this direction a clinician should be able to hear the output of the hearing aid. This can be achieved by any of the following methods:

1. Using stethoclip - This is a simple accessory that allows the clinician to hear the output of the hearing aid without having to wear it. For high powered hearing aids a damper or several dampers can be placed on the stethoclip tubing to decrease the output to comfortable levels for a normal hearing person.

2. A custom earmold (that could fit the clinician's ear) can be attached to a long tube that has an enlarged and flexible cupped end.

3. There are several electronic devices available in which the hearing aid is connected to a coupler and the output of the coupler is amplified and presented through headphones. The advantage of these devices is that a comfortable listening level can be obtained even for high powered hearing aids.

4. For BTE hearing aids that terminates in a dome, the hearing aid with a fresh dome can be worn directly by the clinician.

5. Most real-ear gain analyzers come with a set of headphones that allows the clinician to hear the sounds present in the person's ear canal. Whenever the probe microphone is inserted, the clinician can listen to the sound while the client identifies precisely what aspect of the sound quality is unacceptable. This method is very useful if the clinician has any doubts about the nature of the noise or distortion that the user is describing.

When diagnosing hearing aid faults, it is important to distinguish between noise, distortion and interference.

Image 1.46 showing a stethoclip

Noise:

This is an unwanted part of the output that is present whether or not a signal is being input into the hearing aid. It can originate totally from within the hearing aid and this is known as the internal noise. It can also be an amplified version of some external noise (like that of an air-conditioner). It can also arise from an external non-acoustic source in which case it will be known as the interference.

Interference:

This is creation of a noise in the output of a hearing aid by a magnetic, electrostatic or electromagnetic field near the hearing aid.

Distortion:

This is an unwanted part of the output that is present only when a signal is being amplified. This will usually be audible as a signal of poor quality rather than as something that is present in addition to or in the absence of the signal.

Interference in hearing aids by other electronic devices have been taken into consideration while designing a hearing aid. Common devices that can cause interference is the mobile phone signal.

If the hearing aid is giving intermittent problems then it would be advisable to check the battery and its contacts. If the hearing aid output diminishes in strength / quality each day then returning to good performance in the morning cerumen build up in the wax guard should be a suspect. This is also known as the rain-forest effect because each day the high humidity in the ear canal reactivates and expands the dried out cerumen lodged in the hearing aid.

Presence of wax within the ear canal can cause troublesome feedback.

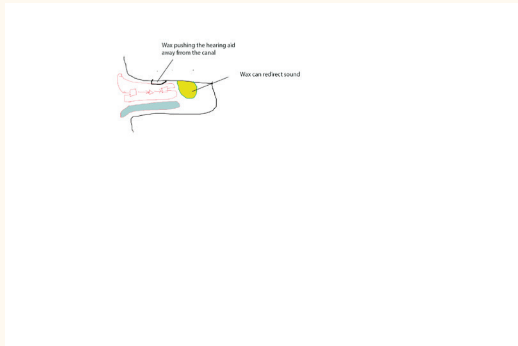


Image 1.47 showing effect of wax on hearing aid efficiency

Feedback is almost always caused by signal leaking within the ear canal. The source of the feedback can be detected with a stethoclip by positioning the open end of the tubing at each of the points where sound could be escaping.

Causes of feed back in hearing aids

Possible cause	Diagnosis	Remedy
Shell improperly inserted	On inspection	User to be taught the correct technique
Shell does not fit snugly	Feed back stops when jelly is applied to the shell	Shell to be remade
Venting insert / Plug has fallen out	Visual inspection	Insert a new venting insert
Microphone / earphone has moved and is touching the casing	Whistling continues when the microphone inlet port is blocked with finger	Return to the manufacturer for repositioning
Microphone tubing detached from microphone	Whistling continues when the microphone inlet port is blocked with finger	Return to the manufacturer for reattachment
Receiver tubing detached from receiver	Whistling continues when the outlet hole is blocked with a finger	Return to the manufacturer for reattachment
Receiver tubing detached from tip of ear shell	Visual inspection	Careful repositioning

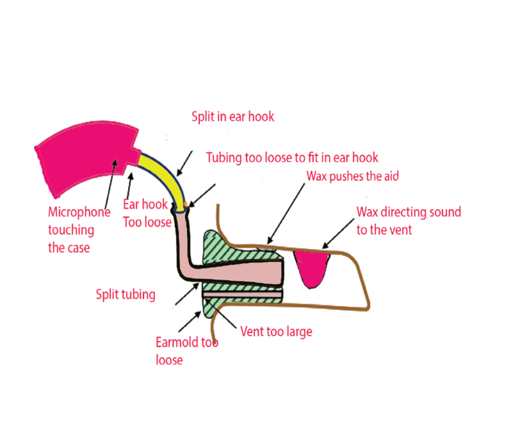


Image 1.48 showing common hearing aid problems

Hearing Aid Earmolds,and coupling

Earmold/Earshell is a pre-molded canal fitting devices. They are collectively known as ear fittings. Earmold is used in Behind the ear hearing aids and earshells are used in intracanalicular hearing aids. Ideally an ear fitting performs three important functions:

1. It couples sound from the hearing aid receiver to the aid wearer's ear canal ia the sound bore (which is a tube). They can affect the gain-frequency response of the hearing aid.

2. It controls the extent to which the inner part of the ear canal is open to the air outside the head (otherwise known as venting). This affects the gain-frequency response and electro-acoustic comfort of the hearing aid.

3. It retains the hearing aid in the ear in a comfortable position.

There are many varieties of ear fitting styles and materials. Some of them could be proprietary to specific manufacturers of the hearing aid. Some of the earmolds could appear very bulky and could completely fill the concha. These molds have a vent drilled through the mold. Other type of earmolds are not bulky and has very little material filling the concha and the canal. These molds are known as CROS molds or Janssen molds.

Ear fittings can be occluding, open or anywhere in between. Occluded fittings are those with no intentional air path between the inner part of the ear canal and the outside ear (residual canal volume). Occluded fittings hence have no vent. The ear fitting completely fills the cross section of the canal for at least part of its length. Occluded ear fittings can have a leakage path around them as a consequence of imprecision in the impression of the ear, imprecision of the mold or shell made fro the impression or flexibility of the ear canal. This leakage path has properties similar to that of a vent and is sometimes referred to as a slit-leak vent.

Open canal fittings are those ear fittings that leave the canal almost completely open for its entire length and is most commonly achieved with an open dome fitting.

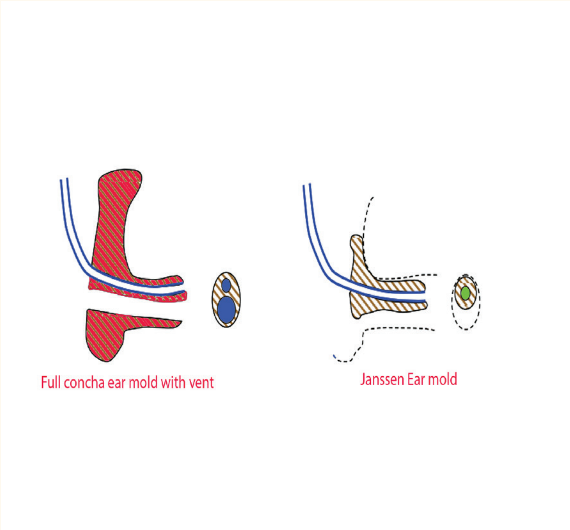
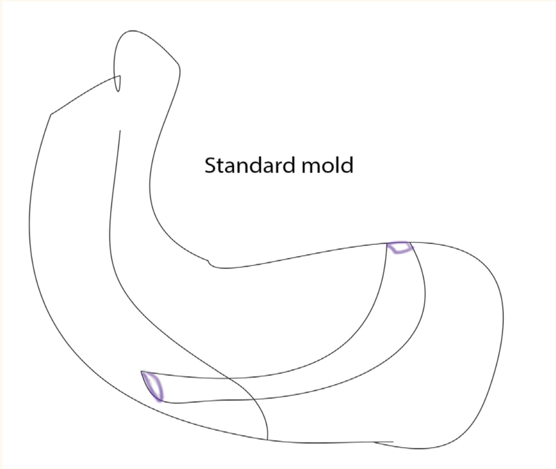


Image 1.49 showing the types of ear fittings

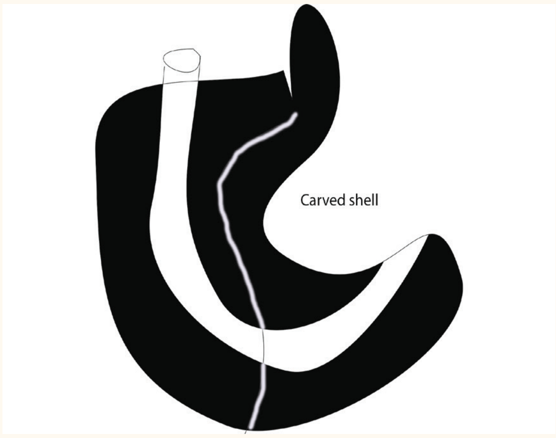
Earmold/Earshell canal fitting styles

Earmolds and shells of different styles fill different portions of the concha and the ear canal. The various parts of the molds and shells can be described by corresponding portions of the ear which they fit. Some anatomical features of the ear have special significance for hearing aid fitting. The inner half of the ear canal which is bony in nature is lined by smooth skin which is about 0.2 mm thick. This area happens to be very sensitive to applied force. The outer portion of the ear canal which is cartilaginous in nature. The skin lining here is thick and is less sensitive. Cerumen is located only in the car-

tilaginous portion of the canal. This area is referred to by hearing aid manufacturers as aperture and the corresponding portion of the earmold is known as the aperturic seal because the mold readily seals the ear canal in this region.



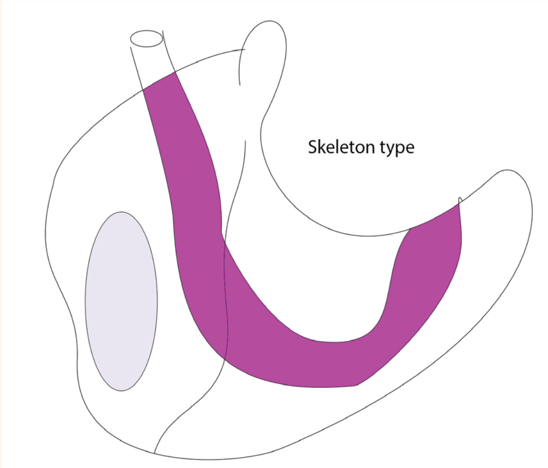
The earmold has two bends which can easily be recognized. The first bend (also the most lateral bend) a pronounced feature on a mold / impression. This is less evident when looking at the ear. The posterior surface of the tragus is continuous with the posterior wall of the canal. The first bend is coincident with the ear canal entrance or a few mm inside the ear canal. The second bend marks the start of the transition from the cartilaginous canal to the bony canal. The first and the second bend are more acute for some people than others. In persons with a sharp first bend, there is also a tendency to have the second sharp bend.

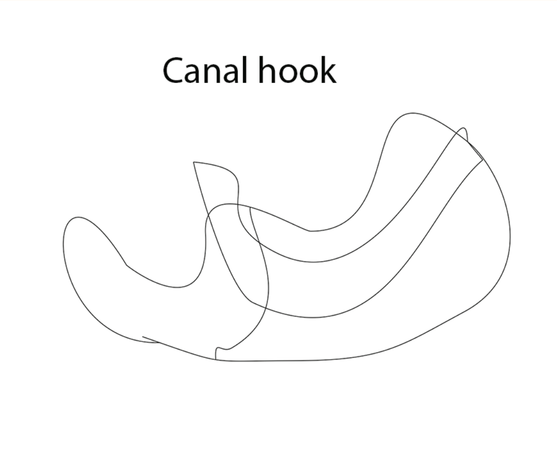


Behind the ear earmold styles

Major difficulties in describing different styles of earmolds is the lack of standardization of names. In fact the American National Association of Earmold laboratories agreed on some names in 1976, many new styles have been designed since then.

Some of the earmolds are given descriptive names (rather useful) i.e. Skeleton.
some of the earmolds were named after the designer who designed it (Janssen).
Some of the earmolds were named after the application in which they were originally used i.e. CROS.





The receiver mold also known as standard or regular mold is the only one that can be used for a body worn hearing aid. A button receiver clips firmly into the ring on the surface of the mold. It can also be used for a BTE aid by clipping a plastic angle piece into the ring. A length of tubing connects the angle piece to the hearing aid ear hook.

The most commonly used fittings are the pre-molded dome shaped canal fittings and thin sound-bore tubing. It is better to ensure that the tubing can easily be replaced by the user is by having an elbow mounted in the earmold to which the tubing can be connected. The sound bore inside the mold consists of a drilled hole instead of a tube. In order to avoid decreasing the high frequency response of the aid, the internal diameter of the elbow should be the same as that of the tubing. The most commonly used fittings are the premolded dome shaped canal fittings and thin sound-bore tubing. These domes have a soft flexible flange and comes in a range of diameters and tube lengths.

These domes are of two types, the open dome with holes in the flange which ensures that the ear canal is left as open as possible and the other being the closed dome without these holes.

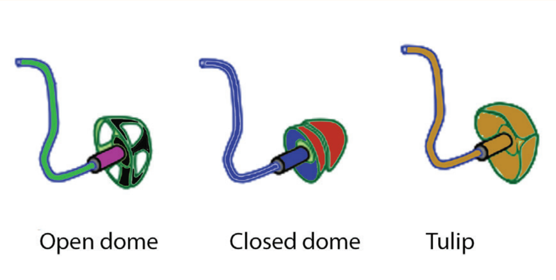


Image 1.50 showing dome type earmolds

The dome and tulip fittings have the advantage of not requiring an ear impression to be made. Same day fitting is possible. Selection of the correct dome size and tubing length is very important. If the dome is too large it would feel uncomfortable, if too small it will fall out or move around. If the

tubing is too long or short then either the hearing aid will not sit comfortably behind the ear or the dome will not sit comfortably within the ear canal. The open dome fittings fulfills the function acoustically as the sleeve mold / vented hollow canal mold.

Some of the earmolds will have helix lock segment intact. These molds can also be ordered with the helix lock removed. Helix lock can also be cut manually / ground by the clinician. The advantage of retaining helix lock is that the mold stays in place thereby maximizing security / safety of thee hearing aid. For this to happen the user should ensure that the user should ensure that the helix lock is properly tucked in under the helix and anti helix. The helix lock by helping the mold to be retained in a correct position decreases the chances of feed back. Users often find it difficult to tuck in the helix lock properly and hence they face the problem of feed back. The helix lock area of the mold is also known to cause pressure discomfort for the user. This is the reason why some clinicians prefer to order molds without helix lock.

Systemic procedures need to be followed for determining how open an ear-mold should be for a particular user. It is also unclear how to systematically choose between molds that differ only in their appearance, fragility, and degree of retention properties. There is no obvious difference in the retention properties or occlusion properties of a shell versus a skeleton, because the material removed to turn a shell into a skeleton comes from the center of the concha region. The general rule being the mold becomes less firmly anchored in the ear as more and more segments are removed from around the rim of the concha, and as the diameter of the canal stalk is decreased below the diameter of the ear canal itself. In users whose pinna moves excessively while talking, chewing, and head turning, the mold or shell would be best retained if it makes minimal contact with the concha, in which case a canal sized ear fitting could be optimal.

Whatever be the style of the ear mold / shell selected, there should be a retention region somewhere in it. A retention region is an area where the earmold / shell pushes against the skin if it were to start moving out of the ear canal. The part of the ear against which the retention region pushes could be thee canal wall, the tragus, anti-tragus or the helix. If the retention region is too small or not sufficiently angled against the exit motion, the earmold will work its way out of the ear. If this retention region is too large/excessively angled against the exit motion, then it will be hard to insert the earmold.

Ear shell styles for ITE, ITC and CIC hearing aids

Full concha ITE

Since the electronics of the hearing aid are inside the shell for these types of aids, the possibilities for alternate shell types within each of these types of hearing aids are fewer. ITE (In The Ear) hearing aids that extend above the crus helix are classified as full-concha ITE's.

Those molds that are fully contained below the crus helix are referred to as half concha ITE's. Those that fit entirely above the crus helix are known as cymba-concha ITE's. These differences are better appreciated only in the lateral view.

If the full concha or half concha ITE's don't extend laterally sufficiently to fill the concha they are referred to as low-profile ITE's. ITE' hearing aids that extend only part of the way along the posteromedial wall of the tragus are referred to as mini-canal hearing aids. These aids can also be thought of as low-profile ITC's.

The distinction between ITE's low profile ITE's CIwCs and deeply seated CICs can be appreciated in the axial section through the ear image 1.51.

The faceplate of an ITE device is approximately parallel to the plane containing the lateral surfaces of the tragus and helix. On the other hand the faceplate of an ITC is approximately at right angles to the posteromedial surface of the tragus. The faceplate of CIC hearing aids may be at the ear canal entrance or medial to the entrance. Any hearing aid that extends to within a few mm of the ear drum is referred to as peri-tympanic / deeply seated. One hearing aid that can be inserted inside the ear canal by the clinician and could stay in place for several months till the battery is depleted is available in the market currently. This is actually a disposable type of hearing aid which can be removed when the battery has got depleted and replaced with another new one. Since the adjustments cannot be performed by the user, it needs to be done using a remote control device.

Earshells used can be occluding or partly occluding. In general the hearing aid becomes less securely anchored in the ear as more of the concha material is removed. CIC hearing aids with little or no material in the concha can be retained in the ear if an appropriate impression technique is used.

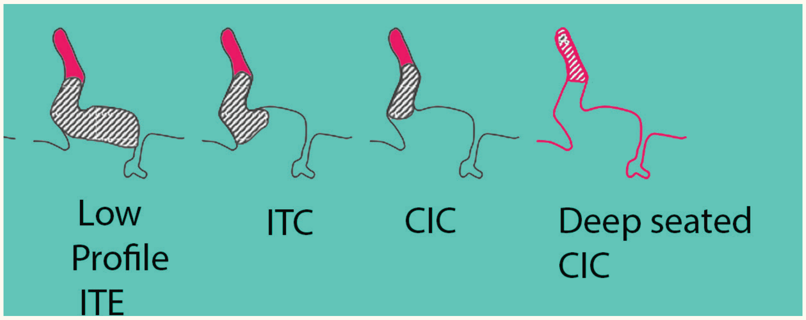


Image 1.51 showing axial view of typical placement of various types of miniature hearing aids

Acoustics of canal fitting

Ear fitting affects three broad acoustic characteristics of the hearing aid which include the shape of the gain frequency response of the ear when it is mounted in the ear, the self perceived quality of the patient's voice and the likelihood of feedback oscillation.

There are three acoustic aspects of the coupling system:

Sound bore

Damping

Venting

These three aspects affect the frequency response in different frequency regions. Sound bore dimensions affect only the mid and high frequency response in different frequency regions (this could be above 1 kHz for BTE aids and above 5 kHz for ITE/CIC aids).

Damping mainly affects the response shape in the mid-frequency region (from 800-2500 for BTE aids and from 1500 Hz to 3500 Hz for ITE/ITC/CIC aids).

Venting mainly affects the low frequency response from 0 Hz up to approximately 1 kHz. If the vent is large enough then it affects the entire frequency range because it leaves the open ear resonance largely intact.

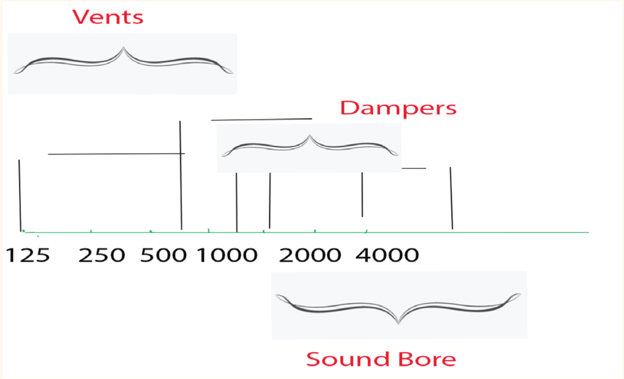


Image 1.52 showing the various frequency ranges affected by the coupling system.

Venting

A good and through understanding of vents, including the venting effects of open canal hearing aids is essential to hearing aid fitting. The vent size is usually selected with the aim of achieving the targeted gain, without the ear canal being excessively occluded, and without the hearing aid oscillating. It is however difficult to achieve these three aims completely. Vents enable an exchange between air in the external auditory canal and the outside atmospheric air. This helps to avoid excessive moisture build up. The venting action enables persons with perforated eardrums to wear hearing aids, provided the perforation is not too large.

Understanding the concept of vents will become easy if the concept of acoustic mass is understood first. A vent is considered to be a column of air surrounded by the walls of a tube. Air like any other substance has mass and hence has inertia. For the vent to transmit sound the inertia should be overcome else air will not be able to move.

Overcoming this inertia is easier at low frequencies than at high frequencies and is much easier for small masses than for large masses. The air column in a vent will not move much and hence will not transmit much sound if the stimulating frequency is high and if the vent has a large acoustic mass. Vents have a high acoustic mass if they are long and narrow.

Real vents are not always tubes with the same dimensions at all points. The concept of acoustic mass helps us to understand how the performance of a vent with a varying diameter differs from that of a constant diameter tube.

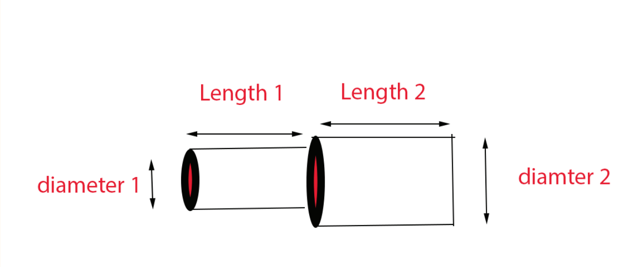


Image 1.53 showing a vent made of two tubes

To calculate the total acoustic mass of a two tube vent as shown in image 1.53 it should be remembered that the value is the sum of the acoustic mass of each segment. In the above image the acoustic mass of the narrow segment will be greater than the acoustic mass of the wider segment. The acoustic mass of vents with more than one diameter as shown in the above figure has a practical application to vents with adjustable apertures and vents that have been widened at one end.

It is really difficult to predict the optimal vent size, the clinician should adjust the vent after a preliminary fitting has been made. One easy way to perform this task is to enlarge the vent diameter by drilling or decrease the vent diameter by filling it with wax / plastic materials. Re drilling can be performed even after filling if needed. Vents can easily be modified quickly if they are ordered along with an exchangeable vent insert plug. These insert plugs all will have the same length (2.5 mm) but differ in the diameter of their internal hole. The different inserts change the acoustic mass of the vent but this could happen only if:

The rest of the vent tube is not so long or so thin that its acoustic mass dominates the total mass.

The leakage around the mold / shell is not so big that the size of the vent is inconsequential.

It should be pointed out that the inserts with the largest and the second largest holes will have almost identical effects because the total vent mass is dominated by the vent tube. Similarly the smallest and second smallest holes will have similar effects to each other. Insert system is still worthwhile since it offers an easy way to obtain two / three different vents. In order to have maximum flexibility in venting then the following factors are important:

1. For the narrowest inserts to be useful, leakage must be minimized by making the mold a tight fit. This is not desirable as it could be uncomfortable and hence not very sensible.

2. For the widest inserts to be useful, the vent tube must be short and wide, which may not be always be possible if the ear canal is narrow.

Effects of vents on hearing aid gains

Vents including leaks and open fittings affect low frequency gain of hearing aids by:

Allowing low frequency sounds out of the ear canal.

Allowing low frequency sounds in to reach the residual ear canal volume without passing through the hearing aid amplifier.

These are two important and separate effects of vents.

Effects of vents on the amplified sound path

When amplified air vibrations emerge from the sound bore into the ear canal, they generate sound pressure into the ear canal. It is actually this sound pressure that is sensed by the ear drum. The smaller the residual volume (the space between the sound bore exit and the ear drum), greater will be the SPL generated. In the presence of an escape route, such as a vent some of the vibrations will leave by that route without contributing to the sound pressure within the canal. The proportion of sound leaving depends on the impedance of the escape route relative to the impedance of the residual canal and middle ear. The vent pathway, having an acoustic mass has an impedance that rises with frequency. The residual ear canal volume, being primarily an acoustic compliance has an impedance that falls as the frequency of the sound increases. The vent becomes an escape route as the frequency of the sound decreases. The vent hence provides a low cut to the frequency response curve.

The extent of the low frequency cut depends on the size of the vent since the vent size determines its acoustic mass.

Effects of vents on the vent transmitted (acoustic) sound path

Vents transmit low frequency sound waves. Sound waves reaching the head will be transmitted directly into the ear canal by a vent. This sound path is non-electronic. The range of frequencies over which the vent transmits sounds into the ear canal without attenuation is the same as the range over which it attenuates sound that has been electronically amplified. Sounds are transmitted into the ear canal without significant attenuation up to the vent Helmholtz resonant frequency. Above this frequency, the vent increasingly attenuates sound directly entering the ear canal from outside the head, so the hearing aid, when turned off, acts like an ear plug.

Real ear occluded gain

This is the sound pressure level in the canal with the hearing aid turned off, relative to the sound pressure level in the incoming field. The sound wave causing SPL reaches the canal primarily via the vent (leakage) path.

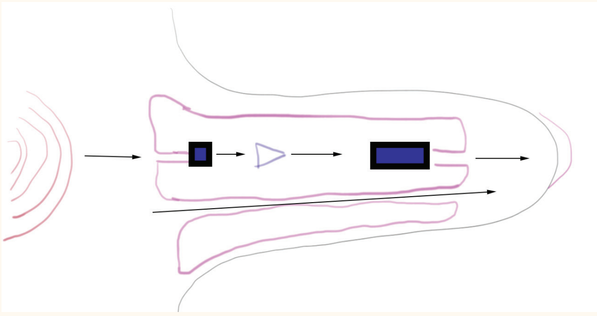


Image 1.54 showing the pathway of sound via the vent of the hearing aid as well as the amplified sound

Real ear occluded gain

This is the sound pressure level in the canal with the hearing aid turned off, relative to the sound

pressure level in the incoming field. The sound wave causing SPL reaches the canal primarily via the vent (leakage) path.

The hearing aid user does not hear either the amplified sound path or the vent transmitted sound path in isolation. As per the diagram above it can be seen that sounds arrive at the ear drum via both routes. The sounds arriving via each path combine in the residual canal volume. Whenever the insertion gain of one path exceeds the insertion gain of the other path by 10 dB or more, the insertion gain of the combined paths is almost the same as the insertion gain of the path with the higher gain. This is because the amount of sound arriving via the path with the lower gain is inconsequential compared to the sound arriving via the dominant path. In the vent transmitted region which can extend up to 1500 Hz in open canal hearing aids, the microphone, amplifier and receiver play no part in the sound received. In the amplified region however, the vent can have an effect if it attenuates part of this region by allowing sound out of the ear canal.

Since vents affect the sound coming out of the hearing aid, they affect the maximum output in the same way as they affect the gain.

Venting and the occlusion effect

When the ear canal is occluded by a mold / shell, users with low-frequency hearing thresholds less than about 50 dB would complain that their own voice sounds hollow, boomy as if they are speaking in a drum. This is known as the occlusal effect. A 2 mm vent is only partially effective in solving the occlusion problem. It decreases the size of the SPL but does not entirely eliminate it. For the user their own voice quality would become acceptable as the mold / shell is more open. A 2 mm vent could be regarded as a very good starting point for fixing the occlusion problem. In majority of cases, the vent will have to be widened to 3 mm before the patient is satisfied with the sound of his / her own voice.

The next way to solve this problem is not to create it in the first place. If the mold / shell completely fills the cartilaginous portion of the canal there will be less occlusion generated sound compared to molds or shells that terminate within the cartilaginous portion. This is of course very difficult to achieve.

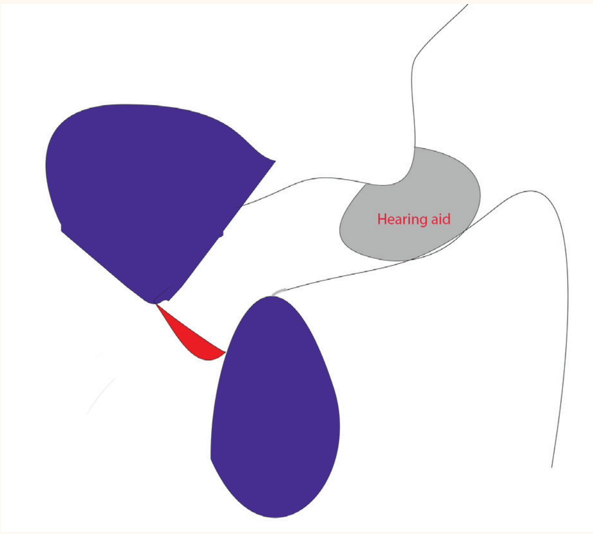


Image 1.55 showing axial view of earmolds that produce a very strong occlusion effect.

Vents and directivity

Accurately designed directional microphones have a directional pattern covering the entire hearing aid bandwidth. This directivity, however will be apparent to the user only at those frequencies where the amplified sound path dominates over the vent transmitted sound path. To maximize the benefits of a directional microphone, the amplified sound path should extend to as low a frequency as

possible. Hence the vent should be made as small as possible.

Since directionality depends on cancellation of the signal picked up at one microphone port by signal picked up at the other port. Vent transmitted sounds can decrease directivity. Even when the aid transmitted sound is 10 dB more intense than the vent transmitted sound, the latter can change the direction at which maximum sensitivity occurs. This can greatly increase the response to rear sounds.

Directivity rapidly reduces as the level of vent transmitted sound comes within 5 dB off the aid transmitted sound. Since the dynamic range compression causes gain to decrease as input increases, aid transmitted sound dominates vent transmitted sound over a wider frequency range at lower input levels than at higher input levels. Hence the frequency range over which directivity is available is least at high levels, which unfortunately is precisely where it is most needed. Vents and open fittings decreases the effectiveness of directional microphones.

Vents and adaptive noise reduction

Similar to directivity, adaptive noise reduction also relies on electronic attenuation of sound at specific frequencies. It applies only to the aid transmitted sound path. Vent transmitted sound therefore creates a level below which sound cannot be attenuated, no matter how poor SNR (Signal to noise ratio) is. Open fittings hence decrease the effectiveness of adaptive noise reduction.

Vents and internal noise

The level of internal noise, like any other amplified sound will be decreased in the low frequencies by vents. Users with near normal low frequency hearing perception of internal hearing aid noise will hence be minimized by making the vent as large as possible.

Parallel versus Y vents

Usually vents are known to cause feed back problems. Another difficulty is fitting them in. This is a problem at the medial end. This is where the Y vent, diagonal vent, or angle vent scores over parallel vents. Y-vent should be avoided unless there is no alternative as it creates problems. High frequency sounds propagating down the sound bore will be partially reflected at the Y junction where the sound bore meets the vent tube. This reflection decreases high frequency gain and also makes high frequency feed back oscillation likely.

If Y vent is absolutely needed then the sound bore and the vent tube should intersect as close of the medial end of the mold as possible. Further the diameter of the sound bore medial to the Y junction should be widened as much as possible. If there is room for extensive widening then there is probably room to avoid the y vent altogether.

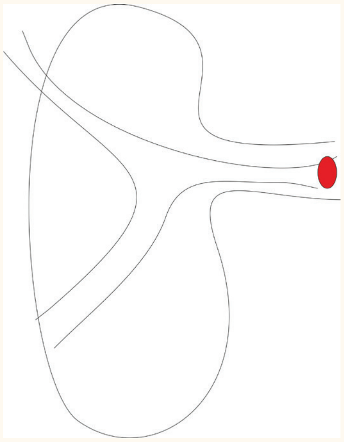


Image 1.56 showing Y vent in position

Open canal fittings, when combined with a thin tubing and a small BTE, gives a very inconspicuous hearing aid. It is more discreet for patients with hair down the top of their ears than even a CIC. The comfort and inconspicuousness of open canal fittings have made them very successful.

These fittings leave the canal sufficiently open that even with the hearing aid turned off, the sound at every frequency reaches the ear drum at almost the same level as when no hearing aid is worn. Since the open canal acts as a large vent, the hearing aid is ineffective in increasing SPL in the ear canal at low frequencies, and is thus limited to providing high frequency amplification. The open canal allows the sound created by canal wall vibrations to escape, so the open fitting successfully avoids the occlusion effect that makes the aid wearer's own voice unacceptably boomy.

Sound bore

This provides the path between the receiver and the residual ear canal volume. Its length is much higher in BTE aids than in other types of hearing aids and hence plays a very important role on the gain frequency response curves of BTEs. There are three types of sound bore systems used in BTE hearing aids. they include:

Earhook BTE

Thin-tube BTE

Rite BTE

The total length of a BTE sound bore ranges from 60-85 mm in adults. For ear hook type BTE's the final 10 to 20 mm of the tubing is contained within the earmold itself. The sound bore in RITE, ITE, ITC and CIC hearing aids is much shorter and contains a tube from 2-10 mm long with a diameter of 1-1.5 mm. The sound bore creates resonances, the frequencies of which are determined primarily by the length of the sound bore, it is also affected by its diameter.

The diameters used in the assessment of resonances happens to be the internal diameter because it is this value that affects the passage of sound along a tube. The thickness of the tubing wall, and hence its outer diameter, affects the leakage of sound through the walls of the tubing. Such a leakage can be a problem in high gain hearing aids, and tubing with extra thick walls is available.

Horns

Horns help in overcoming the impedance mismatch between the impedance of the receiver and the lower impedance values of the ear canal. If the receiver and the ear canal are directly connected together, or if connected via a tube with a constant diameter, much of the power is reflected back from the medial end of the tube rather than being transferred to the ear canal. This can be avoided by gradually changing the diameter of a connecting tube and hence its impedance, there is a more gradual transition from the high impedance receiver to the low impedance canal. This causes less sound to be reflected off the medial end of the tube. This gradual transition is effective only for those frequencies for which the wavelength is less than or compatible with the dimensions of the tube.

Horns with the biggest outlet diameters will give the highest high frequency boost. This boost only occurs for frequencies well above the horn cut-off frequency.

Shorter the length of the horn, the higher will be the cut off frequency. For a horn with an inlet diameter of 2 mm, outlet diameter of 4 mm and a length of 25 mm the horn cut off frequency is

1520 Hz. The boost ideally commences at this frequency and does not reach its full extent until an octave higher than this value. If the horn is made in a stepped manner, the stepped portion has an additional effect. Standing waves will occur within the widened section of the tube because reflections occur at each change of diameter. The quarter wave resonances caused by these reflections can be used to shape and extend the frequency range of the hearing aid.

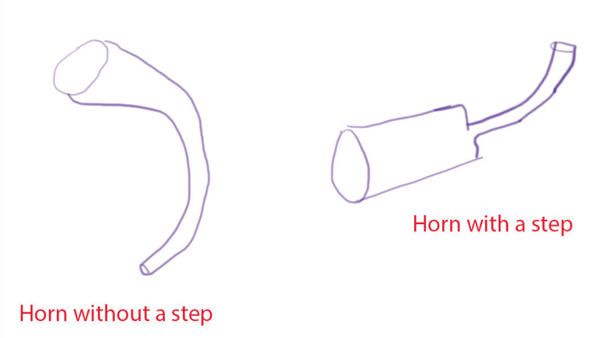


Image 1.57 showing the type of horns

Acoustic horns increase the efficiency with which high frequency power is transferred from the receiver to the ear canal, and hence increases both the gain and maximum output in the high frequency region. This boost is achieved only above a certain frequency which depends on the ratio of the inlet and outlet diameters of the horn and its length. The shorter the horn higher the range of frequencies affected. Horns in BTE fittings provides significant boost at 3 kHz and above, while those within ITEs cannot provide significant boost below 6 kHz.

Horns can be built into BTE earmolds in a number of ways. One simple method is to insert tubing only a few millimeters into the earmold. The outlet diameter of the horn will be determined by the size of the hole drilled into the medial end of the earmold. This method has the following disadvantages:

1. The length of the horn will always be less than the sound bore length of the earmold. The boost may not extend sufficiently far down in frequency.

2. The tubing is poorly retained in the earmold. Glue will have to be applied at the lateral end of the mold, and over time this will cause the tubing to stiffen and crack. Cracking can occur at the point where the tubing is most stressed in daily life.

One alternative to this set up is to use an elbow securely mounted in the lateral end of th e mold to which the tubing is attached. This has the advantage that the tubing can be replaced without having to replace the mold, or without any gluing.

Pre-molded plastic horn like Libby horn can be used. This horn can be used in two ways:

The horn can be fully inserted through the mold for which a bole of approximately 5 mm diameter is required.

In the other method the final 15 mm or so of the horn is cut off and then the remainder of the horn is glued to the lateral end of the mold. Since the mold itself forms the final section of the horn, only a 4 mm hole has to be drilled into the canal portion of the mold. Libby horns are available in 3 mm and 4 mm sizes.

Presence of constrictions have the opposite effect on the horns. They decrease the efficiency with which high frequency power is delivered to the ear canal. They are rarely needed, because hearing loss is greatest at high frequencies and partly because hearing aid receivers become less effective above their primary resonance of 2-3 kHz. Multichannel

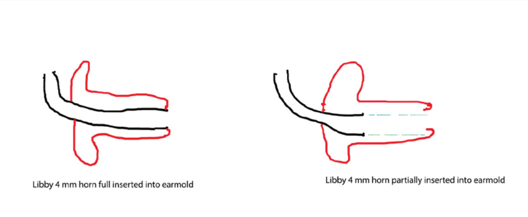


Image 1.58 showing the two types of Libby horns

hearing aids make it easy to decrease gain in specific frequency regions. By combining a constriction with a small cavity achieved by widening the sound bore, an even greater degree of high frequency cut can be achieved.

For closed and minimally vented hearing aids, increasing the length of the canal stalk decreases the volume of the residual ear canal and hence increases the gain at all frequencies. It is a little more for high frequencies than for low frequencies. In open canal fittings, increasing insertion depth increases the gain across frequency, the effect is greatest for low frequencies.

Dampers

These are used to decrease gain and maximum output at frequencies corresponding to resonances in the sound bore. Dampers are most effective if they are placed at locations where the resonance causes the fastest flow of air particles. For wavelength resonances, particle velocity is least at the end of the tube that joins to the receiver. In a BTE, the 1, 3, and 5 kHz resonances are damped effectively as they are moved down the sound bore from the receiver towards the lateral tip of the ear mold. Receiver resonance near 2 kHz is damped slightly more effectively if the damper is placed near the receiver that the tip of the earhook.

Dampers can also be inserted in microphone tubing to decrease the high frequency response of a hearing aid. Dampers are currently not used in thin tube sound bores, out of concern that they could easily get blocked with moisture.

To maximally dampen 1 kHz and minimally damp 2 kHz the damper should be placed as close to the earmold as possible. The tip of the earhook is the most practical position. Dampers placed at the medial end of the earmold would be more effective, but could quickly become clogged with wax and moisture and hence should be avoided.

Procedure to be followed for selecting Earmold and Earshell acoustics

Step 1:

Finding the maximum vent size possible. The target insertion gain is calculated and the appropriate size that would not cause feedback is chosen. For non linear hearing aids, the target gain for low level inputs should be used, as gain is greatest for low level inputs. An allowance of 15 dB should be added if feedback cancellation is available.

Step 2:

Estimating the minimum vent size needed. This estimation should be carried out based on the patient's hearing thresholds at 250 and 500 Hz. This estimate helps in overcoming the occlusion effect. Rule of thumb indicates that low frequency hearing losses greater than 50 dB do not need a vent and low frequency losses less than 30 dB must have at least a 2 mm vent. A 1 mm vent is too small to have any effect on occlusion, the inclusion of such

a vent will make it easier for the clinician to drill / grind a wider vent if it proves necessary.

Step 3:

Deciding on the size of the vent. This will be an easy choice for many patients, as the maximum and minimum vent sizes will be the same, or the maximum vent size will be slightly larger than the minimum. Difficulty will arise for patients with near normal low frequency hearing thresholds and 60-90 dB loss in high frequencies. In this scenario, the maximum vent size will turn out to be less than the minimum vent size. It is worthwhile to order adjustable vents for patients with difficult audiograms.

Step 4:

Selection of sound bore profile.

Step 5:

Selecting a damper. This can be performed efficiently after the hearing aid has been fitted.

Ear impressions

Impression of the external auditory canal needs to be taken before customizing the earmolds and earshells. The entire process of taking an ear impression begins with otoscopic examination.

1. Presence of cerumen should be identified and removed. Presence of cerumen can disrupt the accuracy of the ear impression.

2. Presence of visible signs of outer and middle ear infection is a contraindication for ear impression as it could cause pain.

3. Deep impression should not be created if the ear canal widens sufficiently because removal of the impression will be difficult. Impressions on a patient who have undergone mastoidectomy should be performed only after getting clearance from the surgeon.

Hair in the concha that is long enough to be cut with scissors can be trimmed as it can distort the impression or could get caught in the impression making its removal difficult.

Insertion of a canal block:

This is a small amount of cotton wool or foam that fills the cross-section of the ear canal to prevent impression material flowing deep into the ear canal than required. These are also known as oto-blocks, impression pads, ear dams. The resistance to flow provided by this block enables the impression material to completely fill the canal cross section right down to the desired depth. A piece of strong thread knotted around the block aids easy removal of the canal block.

Blocks can also be custom made or can be purchased in a range of sizes with a pretied thread. The correct size needs to be chosen for insertion. Blocks that are too small may get pushed down the canal by the impression material or could allow the material to flow around the block. Blocks that are too large will not go far enough inside and insertion could be uncomfortable.

The block can be conveniently inserted by pushing it with an illuminated plastic stick. This stick is also known as ear-light/oto-light/light-stick.

The block should ideally be inserted at the level of the second bend.
Mixing the impression material:

Ingredients should be mixed only in recommended proportions. One may be tempted to change the proportion of the ingredients as it could decrease the viscosity, or to make the setting time really short. Adding excessive liquid in a liquid/powder acrylic could make the impression more readily melt or change in shape in heat. Mixing should be through and fast. A spatula should be used to mix the ingredients properly.

The impression material can be scooped into the syringe with the spatula or by pushing the inverted syringe at an angle around the mixing pad.

Filling the ear:

Syringe filled with impression material is partially depressed until the material has started to flow out of the tip. Pinna is pulled upwards and backwards so that the syringe can be inserted as far as possible. If the canal is too narrow syringe extension tips can be used. The piston of the syringe is depressed until the material has covered the syringe tip to a depth of about 6 mm. Plunger is depressed and the tip of the syringe is withdrawn at the rate to ensure that the canal is uniformly filled with the impression material. After the canal is filled, and the concha is nearly filled, the plunger is lowered and the syringe tip is pushed upwards along the back of the concha towards the helix. After the cymba-concha is filled, the plunger end of the syringe is raised towards the front part of the concha close to the tragus. Syringing is finished when the concha is completely filled and is slightly overflowing on all sides.

Marking the impression:
If the hearing aid planned is ITE / ITC and would use a directional microphone, then a horizontal line is scratched across the face of the impression. This line will help the hearing aid manufacturer to position the two ports in the same horizontal plane to maximize the frontal directivity.

Before actually attempting to remove the ear impression a waiting period of 7-10 minutes is a must. Before attempting removal of the impression it should be tested for hardness by indenting it with another sharp object / finger nail. If the indentation created fully disappears then the impression should be considered to be sufficiently cured. If the canal is longer and twisty then a few extra minutes of wait is preferred before attempting removal. Premature removal of the impression could tear it.

Removal of the impression

Before the actual process of removal the patient is asked to open & close his/her jaws a few times. Pinna is pulled down and back. These movements helps to break the bond between the impression and the ear. Firstly the helix part of the impression (helix lock) is extracted. The impression is grasped and pulled out with relevant twisting movements that suits the individual ear.

After removal of the impression the ear canal should be examined to ensure that nothing has been left behind.

Inspection of the impression:

The impression should be inspected for the presence of folds, marks, gaps or bubbles. These blemishes if present in parts of the impression that will be cut away before the product is finished then it

can be allowed. If they are present in other areas then effort must be made to create another fresh impression. Blemish less impression of the canal cavity is a must. The canal block is left attached to the impression. Its angle relative to the impression will give the manufacturer some idea about the direction taken by the canal medial to the end of the impression material. The impression should not be lengthened by adding more impression material after the material h as been removed from the ear. This cannot be done accurately and hence should always be avoided.

In case impression is being performed for hearing aids that needs to extend into the bony canal the following factors need to be assessed:

1. The second bend should be clearly visible in the impression, if not then fresh impression should be taken

2. The impression is inspected under a magnifying glass. The skin lining of the bony canal is smooth and less porous than the cartilaginous canal. This difference in texture can be clearly observed in a good quality impression.

Steps taken before sending the impression to the manufacturer:

Finished impression should be packed carefully in a appropriate container. Distortion of the impression during tansit would be faithfully reproduced in the finished product. Ideally the impression can be scanned with a laser scanner and the image can be transmitted to the manufacturer electronically. This is the currently followed method as it avoids distortion of the impression during transit.

The impression should be annotated with notes being made indicating any abnormalities observed in the ear canal. Hollows in the impression that are caused by bumps in the ear should be marked. If the ear canal is mobile when the jaw moves, the mobile region should be marked in the impression. Or ideally the impression should be taken with the jaw open.

Ear impression techniques for CICs and high gain hearing aids:

CIC hearing aids and high gain hearing aids may require earshell / earmold to fit the ear canal more tightly than is necessary for other hearing aids. This tight fit is needed to avid feedback oscillation / in the case of CIC to retain the hearing aid in the ear.

CIC hearing aids are retained in the ear by the bends in the ear canal and by the variations in cross-sectional area and shape that occur along the axis of the ear canal. The widening of the ear canal that occurs at the second bend is rather important. It is hence essential, therefore that impressions for CIC hearing aids be sufficiently deep to include this second bend and should preferably extend at least 5 mm past the second bend. When the canal block is inserted this deep, the impression material would expand the cartilaginous canal along its entire length providing a more secure fit.

CIC hearing aids and to some extent ITC hearing aids may also require a good fit to ensure that the hearing aid does not work its way out of the ear. Movement of the hearing aid can occur because the ear canal changes shape when the patient moves the jaw. As the jaw opens, the condyle of the mandible moves forward and this pulls the anterior wall of the canal forward. The width of the ear canal increases by 10% for a jaw opening of 25 mm measured between the upper and lower incisors. In the presence of ill fitting dentures, or the patient

with a temporo mandibular joint disorder then the jaw could over-close and the variation in the canal size could even be still more. The best solution of an excessively mobile ear canal is to take the ear impression with the jaw held open by using a 25 mm spacer. Studies have revealed that the CICs made from open - jaw impressions to be as comfortable as those made from closed - jaw impressions.

Another way of ensuring a tight fitting earmold / shell is to progressively build up the size of the impression in the ear of the patient. The three stage impression technique uses ear impression material with different viscosities to make a tight fitting and accurate impression oft he ear canal. It should be pointed out that the open jaw technique is fairly simple and faster.

Steps performed to produce a tightly fitting and comfortable earmold, or earshell:

1. The impression should be taken with the jaw open.

2. Two/three stage impression of the ear canal is taken

3. A special build-up during earmold construction is requested.

4. A viscous non-runny silicone impression material is used.

5. The impression is patted down before it hardens.

Deep insertion CICs require really deep ear canal impressions. The cotton block used for this purpose should be located at / within a few millimeters from the ear drum. The cotton block designed for this purpose uses cotton only for about 2 mm depth and is securely attached to the end of a hollow plastic tube. During extraction of the impression, the tube allows the pressure medial to the impression to equalize to the ambient air pressure there by minimizing pain during removal. Since the cotton block is close to the ear drum, the impression material used should be of low viscosity to minimize the pressure it exerts against the eardrum during injection of the impression. The medial surface of the cotton block can be coated with lubricant prior to the insertion for making the removal fairly easy.

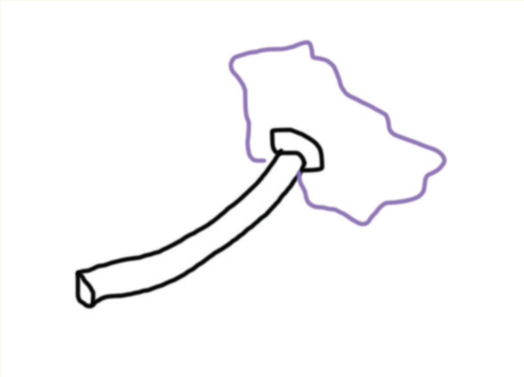


Image 1.59 showing Thin cotton block with attached pressure relief tube used to take very deep impressions

Types of materials used for ear impression materials:

Three types of materials are used for taking ear impressions. If two materials are mixed they undergo a chemical reaction.

Acrylic material - This is usually mixed by combining a liquid and a powder.

Condensation-cured silicone material - This is mixed by combining two pastes.

Addition-cured silicone material

Ear impression materials should have certain of the

following properties to ensure designing a comfortable earmold / earshell.

Viscosity:

Low viscosity materials are easy to syringe and are least likely to expand / distort the ear canal. They are recommended for making impressions for CIC hearing aids in order to make a faithful reproduction of the ear canal.

Dimensional stability:

If the ear impression shrinks in hours / days following its making, the earmold / earshell will also be smaller to the same degree unless some compensation build-up is applied during the manufacture of the earmold or shell.

Stress relaxation:

When force is applied to an impression to remove it, it stretches, compresses and twists as it is pulled through the beds and tight parts of the ear canal. After removal, it is desirable that the impression spring completely back to the size and shape of the ear canal. The extent to which this happens is called stress relaxation. Silicone materials have good stress relaxation properties and acrylic materials do not.

Hardness:

Softer impression material is flexible when set and hence easire to extract from the ear than harder impression material. Viscosity has got nothing to do with hardness and should not be confused with hardness.

Release force:

Ear impression materials are designed to conform closely to minute variations in the surface of the ear and the ear canal. This makes the impression to adhere to the skin and to make the removal smooth a release agent should be built into the impression material. This produces an oily feel of a completed impression.

Construction of Earmolds / Earshells:

Earmolds and earshells are commonly referred to as otoplastics. They are made by the hearing aid manufacturer, from the impression obtained by the clinician. The impression can also be considered to be the negative of the ear. There are two methods of earmold/earshell construction: the investment method and the computer aided manufacture method.

Investment method:

The impression is placed into liquid silicon or other material that cures around it to make a positive copy of the ear. This positive copy is also known as the investment, and the finished mold or shell is made from this investment. A mold is made by filling the investment with liquid mold material and allowing it to harden, often accelerated by ultraviolet light. A hollow shell is made in the same way, but for the fact that most of the mold material is poured out before it hardens, leaving just a thin shell covering the inner surface of the investment. For ITE/ITC/CIC hearing aids, the manufacturer trims the shell to the desired size, inserts the electronic and mechanical parts, attaches the faceplate, and sends the complete hearing aid back to the clinician.

Computer aided manufacture:

The starting point for CAM is a standard impression which must be made of opaque material (silicon). Instead of making an investment the manufacturer scans the impression with a laser to

produce a numerical representation of the three-dimensional surface, and hence of the parts of the otoplastic that contact the ear in fine detail. For custom devices, the manufacturer using an automated program can insert in virtual reality (computer screen) a representation of the hearing aid components (receiver,and faceplate with attached battery compartment, microphone and switch or control). The fit can then be confirmed virtually for the fit and then the shell is finalized.

When all the dimensions of the shell have been specified, the corresponding numerical values are transferred to the plastic printer. This data is used to control lasers that cause a bath of light sensitive liquid plastic to polymerize (solidify) in just those positions where the shell or earmold is to be formed. This process goes by the name stereo lithography. An alternative to this process is the laser sintering which builds up the shell or earmold by melting nylon powder,which then solidifies in place, in only those positions where solid material is needed.

The property that affects the comforts and acoustic performance of an otoplastic is its hardness. This value can be measured by noting how large an indentation occurs when a standard cone / ball shaped object is pushed into the material by a standard force. Sharp indentors and large forces are used for hard materials; blunt indentors and small forces are used for soft materials.

The tool used to measure this is the durometer, and the resulting indentations are expressed as numbers between 0and 100 on a shore hardness scale. Larger this value harder is the material. There are many scales used for measuring these values.

Scale A is suitable for softer otoplastic material.

Scale D is suitable for harder material.

A reading of 90 on A scale is approximately equivalent to a reading of 39 on the D scale.

Soft materials are intrinsically more flexible than hard materials. Greater flexibility makes earmolds easier to insert in a tortuous ear canal and may make them more comfortable when the ear canal changes shape. Soft materials also provide a better seal to the ear as they are not polished during manufacture.

Otoplastics could contain more than one material. Commonly this comprises of a soft material in the canal stalk / or in just the deepest part of the canal stalk, combined with a hard material in the more lateral parts. Advantages of soft material like superior retention, feedback, and comfort properties can be combined with the superior durability of the harder material surrounding the lateral parts of the hearing aid. A potential problem facing these mixtures could be the possibility of fractures occurring at the plane where these two material join.

Advantages of the soft materials should be weighed against the greater deterioration of soft materials with time. The key requirement for comfort is to have some amount of flexibility at the interface between the otoplastic and the ear. If the user's ear is sufficiently soft and flexible, which is very common in old individuals, soft material need not be used in the otoplastic. Hard materials can be safely used in these patients.

Skin lining of the external auditory canal could react to an otoplastic. This reaction could be caused by an allergic reaction to the specific material or may be the result of prolonged occlusion no matter what material is used. One of the common causes for this allergic reaction could be that a small pro-

portion of the original monomer did not cure into a polymer when the earmold was constructed.

Solutions that can be used to overcome this problem include:

1. Using an otoplastic that has been heat cured rather than cold cured as heat curing decreases the proportion of uncured monomer.

2. Trying an otoplastic based on a different, low allergenic chemical such as silicone / polyethylene.

3. Gold plating the otoplastic

4. Referring the patient to contact allergy test

5. Trying a open mold

6. Alternating hearing aid use between ears

7. Use of bone conduction aid if all these efforts fail.

Instant earmolds and hearing aids:

An earmold sometimes could be needed instantly for purposes of demonstration or as a temporary solution while awaiting a repair.

1. Using a hearing aid designed to be fitted with a stock canal fitting, typically comprising of a thin pre-bent tube (which can be chosen from amongst 4 different diameters) is attached.

2. A temporary earmold can be formed from a foam plug with a tube going through it, which can be coupled to an elbow and a tube. These provide a better seal than a conventional custom ear mold and are more comfortable.

3. Custom earmold can be made in minutes by taking an impression using a two-part silicone material. The resulting impression after some trimming could be used as the final earmold.

4. A modular prefabricated ITE, ITC or CIC hearing aid of appropriate size can be chosen.

Modifying and repairing earmolds and earshells

Common reasons for modifying or repairing earmolds and earshells are to:

1. Remove helix locks to ease insertion

2. Shorten / taper canal stalks to ease insertion

3. Remove material from the inter-tragal ridge, conchal rim or the canal stalk to eliminate pressure points.

4. Widen / shorten vents to decrease occlusion

5. Constrict vents to decrease feedback

6. Thicken canal stalks to decrease feedback

7. Replace loose / hardened tubing

Earmolds and earshells can be modified in the clinic using suitable tools and materials. For the BTE earmold, a hand-held motor tool is adequate. to re-obtain the high luster usually seen in ITE/ITC/CIC hearing aids, buffing & polishing wheels are needed.

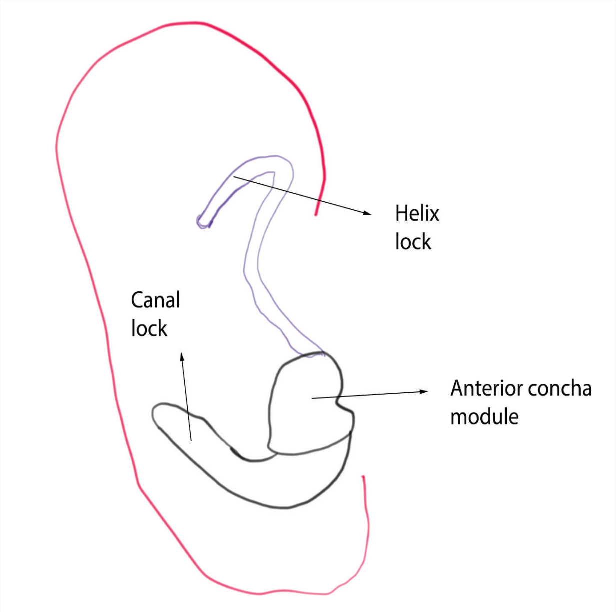


Image 1.60 showing helix lock and canal lock

Modifying vents:

Earmolds / earshells are made less occluding by enlarging the diameter of the vent, shortening the vent length or a combination of both. Vent diameter could be enlarged by drilling / grinding. Vent length is shortened by grinding away the mold / shell from either end of the vent. Before modifying any custom earshell, a strong light is viewed through it to identify the location of the components and to estimate the thickness of the shell walls. The shell should also be checked for the presence of a poured vent (molded vent) encased in solid plastic, rather than just a vent made of tubing.

Re-tubing earmolds:

Replacing the tubing of BTE earmold is commonly needed and is fairly easy to perform. If needed the existing hole can be reamed out with a drill bit. A pipe cleaner dipped in solvent can be used to remove any old glue or debris. To facilitate easy insertion the end of the new tubing should be cut at an acute angle. Unless the new tube is an excessively tight fit in the earmold, the new tubing can be pushed into the earmold.

If the earmold is made from acrylic, the tubing should be glued in. Glue should be applied completely around the perimeter of the tube so that there is no crack for cerumen to penetrate. If the earmold is made from silicone, the tubing must be held in place mechanically. A collar is built into the tubing prior to the tube being inserted. Some collars are designed to slide in only one direction, other types slide more freely over the tubing and must be glued in place.

Material needs to be added to earshells if a grinding operation breaks through the wall of the shell, thus exposing the inner cavity and the electronic components. Material can be added by brushing plastic build-up material on to the earshell. Another reason for adding material either to the shell of mold is to prevent feedback oscillations.

Compression system in hearing aids

The major role of compression in the hearing aid is to decrease the range of sound levels in the environment to better match the dynamic range of a hearing impaired person. The compressor that achieves this reduction should be most active at low, mid, or high sound levels. Commonly it can vary its gain across a wide range of sound levels, in this case it is known as a wide dynamic range compressor. Compressors can be designed to react to a change in input levels within a few thousandths of a second or their response can be made so gradual that they could take many tens of seconds to fully react. This difference in compression speeds are best suited for different types of people.

The reaction of the compressor can be depicted on an input-output diagram o on an input-gain diagram. The compression thresholds, which is the input level above which the compressor causes the gain to vary are clearly visible on these diagrams.

The simple compression systems are classified into:

Input controlled - The compressor is controlled by a signal prior to the hearing aid's volume control.

Output controlled - The compressor is controlled by a signal subsequent to the volume control. This can prevent the hearing aid causing loudness discomfort, or the signal being peak clipped.

Fast-acting compression with a low compression threshold can be used to increase the audibility of the softer syllables of speech, while the slow acting compression will leave the relative intensities unchanged, but would alter the overall level of the speech signal.

This classification is not relevant for hearing aids with no volume control and are inadequate for hearing aids with multiple, sequential compressors.

Compression applied with a medium compression threshold will make hearing aids more comfortable to wear in noisy environments.

Multichannel compression can be used to enable a hearing impaired person to perceive sounds with the same loudness that would be perceived by a normal hearing person listening to the same sounds. Compression can also be used to decrease the disturbing effects of background noise by reducing gain most in those frequency regions where the SNR is poorest.

Intact human auditory system has an immense ability to hear a wide range of sounds in the environment, from soft to intense sounds, average conversational speech of course falls comfortably in the middle of this dynamic range. Sensorineural hearing loss reduces this dynamic range of hearing available. Hence the individual is unable to hear soft sounds, average conversational speech is barely audible, and intense sounds are heard as loudly as the normal ear.

Role of compression:

The main role of compression is to decrease the dynamic range of signals in the environment so that all signals of interest would fit within the restricted / reduced dynamic range of the hearing impaired individual. The intense sounds will have to be amplified less than weak sounds. Compressor can be considered as an amplifier that automatically reduces its gain as the signal level somewhere within the hearing aid increases. There are many ways in which the gain can be varied to decrease the dynamic range of the signal.

Type I:

In this type the gain starts to reduce as soon as the input level raises above weak. By the time a moderate input level has been reached, the gain has been sufficiently reduced. Linear amplification can be used to for all higher input levels.

Type II:

This approach is opposite to that of the above one. Here low level sounds are amplified linearly, but the inputs from moderate to intense sounds are squashed into a narrower range of outputs.

Compression limiting:

This is an extreme case of compression where the output is not allowed to exceed a set limit.

Wide dynamic range compression:

Here compression is applied more gradually over a wide range of input levels.

A compressor is a dynamic device and its job is to change the gain of the hearing aid depending on changes in the level of the audio signal. When the output level first rises, the detector starts to pass on the increased level to the compressor control circuit. As a first step the detector first has to convert the waveform to a smooth control signal. This involves rectification and then smoothing. The result of this smoothing is that following an increase in signal level, the detector output increases gradually to its new value. During the time taken for this process to occur, the compressor is not aware of the full extent of the increased signal level and hence it does not turn the gain down sufficiently to compensate for the increase. The amplifier hence initially passes the increase without compression, until the compressor reacts to the new input level. The time taken for the compressor to react to an increase in signal level is referred to as the attack time.

Attack time is defined as the time taken for output to stabilize to within 2 dB or 3 dB of its final level after the input of the hearing aid increases from 55-80 dB SPL. Eventually the compressor fully reacts to the increased signal level, i.e. its gain has been decreased compared to its previous gain.

A similar event occurs when the input signal decreases in level. The detector progressively reacts to the new input level so for a while the compressor amplifies the low level signal with the gain that was appropriate to the high level signal preceding it. The control signal decreases gradually and consequently the gain and output signal increase gradually. The release time is the time taken for the compressor to react to a decrease in input level. The attack and release times can be reduced to short values (even zero), the consequences are not desirable. The attack and release times have a major impact on how the compressors affect the levels of the different syllables of speech.

Compressors intentionally change the signal's envelope while leaving the fine structure unchanged. Signal envelope is an imaginary line drawn through the extremities of a wave form. the envelope gives an indication of the level of a signal, without showing the fine structure of the wave form.

There is no necessity for a compressor to have a single attach and release time. The release time and attack time should depend on the signal being amplified. Rapid attack and release is best for protecting the aid wearer against brief intense sounds. It should be stressed that a rapid increase in gain during the pauses in speech could cause greater gain to be applied to the background noise that to that of speech.

Currently available hearing aids have an adaptive release time. In this the release time is short (20 ms) for brief intense sounds, but becomes longer (1 sec) as

One compressor which uses a combination of fast and slow acting detectors when used at the input of a hearing aid has been referred to as a dual front end compressor. This has demonstrated to have the advantages expected of adaptive release time compression.

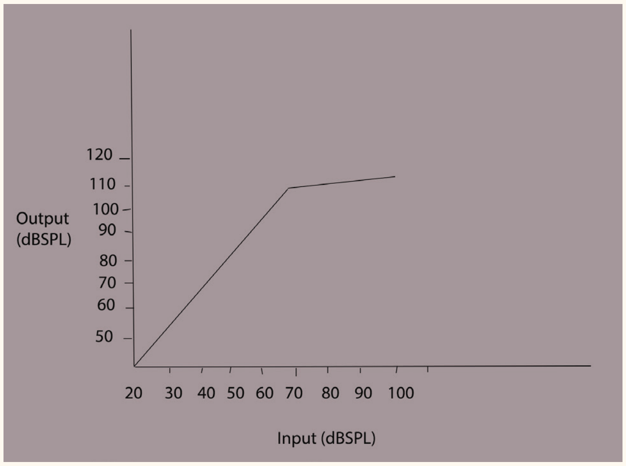


Image 1.61 showing the typical I/O curve of a hearing aid.

The I/O curve shows that the output of the hearing aid does not exceed 110 dBSPL.

The duration of an intense sound increases. When an adaptive release time is combined with a short attack time, a brief intense sound will cause the gain to rapidly decrease and then rapidly increase when the intense sound ceases. This rapid action provides the much needed protection against excessive loudness for brief sound without affecting the audibility of following sounds. Adaptive release times can be achieved by using a single detector with properties that vary with the signal, by controlling a compressor from multiple detectors, or by using multiple compressors in succession.

Essential terminologies that needs to be understood

ANSI:

American National Standards Institute (ANSI) has created ANSI 3.22. The ANSI 3.221 is the hearing aid standard that has been in use in US since 1977. This standard defines the terminology around hearing aids and how manufacturers test their hearing aids.

Input:

This refers to the acoustic signal that enters the hearing aid. ANSI defines input level as the sound pressure level at the microphone opening of the hearing aid. Input level is expressed in dBSPL.

Output:

This refers to the amplified signal that is delivered to the ear. The output level is expressed in dBSPL.

Input/Output Function:

This is a graphical representation of the output of a hearing aid at various input levels. As per ANSI standards the I/O graph has the output SPL on the Y axis with the input on the X axis. The scales for both axis should be linear and of equal spacing. I/O curves are run at individual frequencies.
Gain:

The refers to the amount of amplification applied to the input signal. ANSI defines gain as the difference between the output SPL in a coupler and the input SPL. Gain is expressed in dB. the formula for calculation of gain is:

Gain = Output - Input

As per the I/O curve shown above the input of 40 dB causes an output of 70 dB making the gain value to be 30 dB.

Input / Gain Function:

This is a graphical representation of the gain of a hearing aid at various input levels.

Frequency Response Curve:

This is a graphical representation of the hearing aid output as a function of the frequency. The input level and overall gain of the hearing aid are fixed when the frequency response is measured.

The currently available digital hearing aids has demonstrated some new possibilities for controlling compressors. Overshoots can be fully avoided if the compressor is able to detect and decrease its gain before the signal level increases. This is possible if the signal could be delayed by a few milliseconds allowing time for the detector to fully react to the signal before it could reach the compression amplifier. This feature goes by the name look-ahead compression. A compressor can also operate with a feed-forward control circuit instead of feedback control circuit.

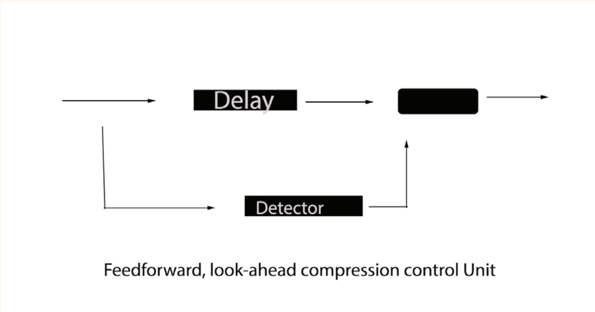


Image 1.62 showing a block diagram of a feed-forward, look-ahead compression control uni

Creative application of adaptive time constants already available in some commercial hearing aids is the concept of adaptive dynamic range optimization (ADRO). This algorithm is a compression technique that applies multiple time constants and several rules and rules in a sequential order to set the gain at each frequency so that:

1. Gain is reduced to avoid the maximum level of the signal exceeding loudness discomfort level.

2. Gain is reduced to avoid the upper levels of speech which happens to be the 90th percentile exceeding the comfortable range.

3. Gain is increased to avoid lower levels of speech (30th percentile level) becoming inaudible

4. Gain is never allowed to exceed a predetermined maximum value aimed at avoiding feedback oscillation and excessive amplification of background noise.

The first rule is applied with a short attack time of around 20 ms, whereas rules 2 and 3 are applied with a very long attack and release times, respectively of around 7-10 seconds. Application of these multiple rules affecting the gain at each instant in each narrow frequency region is described as fuzzy logic.

Attack time and release time interact to affect the signal level at the output of a compression hearing aid. At one extreme, very short attack times combined with very long release times detect close to the peak level of the signal. On the other hand if the attack time and release times are equal, the level of the signal detected is closer to the mean level of the signal. Since this level is much lower than the peak level, the compressor will think that the signal is weaker, and hence would cause great-

er amplification than if the peak levels controlled the compressor. On the other hand reducing the release time without changing attack time would cause output levels to increase by several decibels. The greater the compression ratio the greater will be the effect of attack and release times on output levels.

The attack and release times reveals how quickly a compressor operates. when the gain changes alone are measured it is assumed that the compressor has had time to fully react to variations in signal level.

The sound pressure level above which the hearing aid begins compressing is referred to as the compression threshold. Most hearing aids amplify linearly for input levels immediately below the compression threshold. Compression threshold is defined as the input SPL at which compression commences. The onset of compression can actually be very gradual.

Static compression characteristics

The attack and release times indicate how quickly a compressor operates. Different terms are used to indicate how much a compressor decreases the gain as the level rises. When these gain change values are measured, it is safely assumed that the compressor has had the time to fully react to variations in signal level. Hence the term static characteristics are applicable to signals that are longer than the attack and release times.

The sound pressure level above which the hearing aid begins compressing is known as the compression threshold. Most hearing aids amplify linearly for input levels immediately below the compression threshold. The term compression threshold is defined as the input sound pressure level at which compression commences. Standards of measurement defines compression threshold as the point at which the output deviates by 2 dB from the output that would have occurred had linear amplification continued to higher input levels. Once the input becomes sufficiently intense that forces compression to commence the gain decreases with further increases in the input levels. The term compression ratio describes indirectly how much the gain decreases. Compression ratio is defined as the change in input level needed to produce a 1 dB change in the output level.

Static characteristics apply only to signals of long duration. Hearing aids act in an increasingly linear manner when the intensity fluctuations become increasingly rapid. Hence for rapidly changing signals, the effective compression ratio is less than the static compression ratio. Phonemes and syllables vary widely in duration, but it is sensible to ascertain what the effective compression ratio might be for a signal with a typical syllable duration of about 120 ms. Only when the attack and release times are much less than 120 ms will the effective compression ratio equal the static compression ratio.

Role played by location of volume control in a hearing aid

The location of volume control relative to the compressor plays a vital role in the signal output when the user twiddles with the volume control. If the compressor precedes the volume control in the signal chain the effect of volume control on the compression characteristics is none. Compression in these hearing aids commences at the same input sound pressure level for all settings of the volume control. The volume control simply determines the size of the output signal. This particular setup is defined as the input-controlled compressor / automatic gain control.

If the volume control is situated in front of the compressor then it affects the signal before it could reach the compressor. This setup is also known as the output-controlled compression. If the volume control is turned down, the amount of signal that reaches the compressor would no longer be enough for compression to commence.

In hearing aids with wide dynamic range compression facility, it is not common to have a volume control. In these hearing aids, the main distinction between input and output control literally disappears.

Multichannel compression:

Multichannel hearing aids splits the incoming signal into different frequency bands and each band of signal passes through a different amplification channel. In a multichannel compression hearing aid, each channel contains its own compressor.

Reasons for using different compressors for different frequency regions include:

1. The amount of compression varies with hearing loss, but hearing loss varies with frequency

2. The amount of compression varies with signal level, but signals and noises in the environment have more energy in some frequency regions than in others

Multichannel compression enables this variation of compression with frequency to be achieved. Greatest amount of compression would occur if the compression ratio is high and the compression threshold is low.

In a single channel compression hearing aid, when the compressor turns down the gain, signal components at all frequencies are decreased in level. It may not be appropriate to have signal components at one frequency being attenuated just because there is a strong signal at another frequency. Multichannel compression avoids this problem.

Compression can vary from one channel to the next. The degree of compression often either increases or decreases with frequency.

A classification scheme has been evolved to describe this overall behavior. When the degree of compression is greater in the high frequency channels compared to low frequency channels, there will be a greater high frequency emphasis at low input levels than at high input levels. This feature is labeled as a Treble increase at low levels (TILL) response. On the other hand when the degree of compression is greater in the low frequency channel than in high frequency ones then there will be less high frequency emphasis at low input levels than at high input levels. This feature is labeled as a Bass increase at low leevels (BILL) response. These terms have literally been forgotten these days since compression in a multichannel hearing aid has become more complex.

Avoiding discomfort, distortion and damage:

The hearing aid output cannot be permitted to keep increasing in level as the input level to the hearing aid increases. If excessively intense signals are presented the hearing aid wearer will feel the discomfort. The aid wearer's loudness discomfort level provides an upper limit to the hearing aid amplification. Excessively intense signals could cause damage to the aid wearer's residual hearing ability.

These peaks can be limited by either peak clipping or compression limiting. Compression limiting is preferred over peak clipping because peak clipping creates distortion. Waveform distortion created by peak clipping is far more troubling than the envelope distortion created by compression limiting.

When compression limiting is used to control the hearing aid output, then it must be an output-controlled compressor, or else the output would rise and fall with the position of the volume control. This scenario may not be acceptable. A high compression ratio is needed, so that the output SPL does not rise significantly for very intense input levels. The attack time should also be short so that the gain decreases rapidly enough to prevent discomfort caused by loudness of sound. This gain reduction should also be removed rapidly so that sounds following an intense sound are not overly attenuated, hence the release time should also be very short.

If a hearing aid does not include a compression limiter, peak clipping will occur once the input signal becomes sufficiently intense. If the hearing aid is provided with a wide dynamic range compression, the input level needed to cause peak clipping could become so that peak clipping very rarely occurs.

Reduction of inter-syllablic and inter-phonemic intensity differences:

Most intense speech sounds (some vowels) are about 30 dB more intense than the weakest sounds. In persons with very reduced dynamic ranges it may be difficult to achieve and maintain a volume control setting that makes the weakest sounds of speech sufficiently audible to be understood without the most intense sounds becoming excessively loud. This problem can be addressed by including a fast acting compressor that increases its gain during weak syllables / phonemes. This type of compression is known as syllabic compression or phonemic compression.

Compressors intended to decrease the intensity differences between syllables must have compression thresholds low enough for the compression to be active across the range of short-term input levels that apply to speech. They must have compression ratios high enough to significantly decrease dynamic range, but low enough to leave some intensity differences intact.

Compression parameters used to reduce inter-syllabic level differences:

Input controlled compression

Compression ratio >1.5:1, but <3:1

Attack time from 1-10 ms and release time from 10-50 ms

Compression threshold <50 dB SPL

Single or multichannel

Ideal compression settings to decrease long-term level differences:

Input controlled compression

Compression ratio > 1.5:1, but < 4:1

Attack time > 100 ms and release time > 400 ms

Compression threshold <50 dB SPL

Single or multichannel

Role of compression in increasing sound comfort:

Compression limiter could be expected to solve problems caused by excessive loudness since people may not like the signal being close to discomfort level for a large proportion of the time. It may not be satisfactory to simply reduce the intensity of signal from the hearing aid as it could decrease the usable dynamic range by a greater degree than the person's hearing loss. This problem could be solved by a form of compression that is more gradual than compression limiting.

Compression settings to increase sound comfort:

Input controlled compression

Compression ratio > 1.5:1, but < 4:1

Attack time and release time unknown, not important. Release time should not be too short.

Compression threshold approximately 65 dB SPL

Single / multichannel

Normalizing loudness:

The most common approach to deriving compression characteristics is to normalize the perception of loudness. The principle of loudness normalization is simple i.e. for any input level and frequency, give the hearing aid the gain needed for the user to report the loudness to be the same as that which a person with normal hearing would report.

Loudness can only be measured subjectively. Currently the most popular way is to ask the hearing impaired person to rate the loudness is to ask the hearing impaired person to rate loudness using some special terms. These terms are included under categorical scaling of loudness. These scales commonly have about seven different labels.

Maximizing intelligibility:

Multichannel compression can be used to achieve in each frequency region the amount of audibility that maximizes intelligibility, subject to some constraint about the overall loudness.

Noise reduction:

The interference caused by background noise is the biggest problem faced by hearing aid wearers. Compression is used to decrease the effects of noise. The principle behind this is as follows:

1. Noise usually has a greater low frequency emphasis than speech.

2. Low frequency portions of speech are most likely to be masked so little speech information may be available at low frequencies.

3. The low frequency portions of noise may cause upward spread of masking and so masks the high frequency portions of speech.

4. The low frequency portions of noise contribute most to the loudness of the noise

5. Signal to noise ratio generally decreases as the sound pressure level in the environment decreases.

Compression settings for noise reduction:

Gain reduction where SNR (signal to noise ratio) is worst.

Sometimes approximated by compressing only the low frequencies.

Attack time and release time long or short.

Compression threshold medium.

Usually implemented by multichannel signal processing.

Compressor combinations in hearing aids:

Hearing aids can contain more than one compressor. Ideally a hearing aid could combine:

1. An input compression limiter to prevent very high input signals from overloading the circuitry inside the hearing aid. Many hearing aids include this feature.

2. A slow acting compressor to decrease the dynamic range associated with changes in long-term input level or alternatively a multichannel structure with a slow acting compressor in each channel and

3. A fast acting output controlled compression limiter to prevent output from exceeding the required maximum output limit without waveform distortion.

Advantages and disadvantages of different compression systems:

Advantage of a compression system depends on:

1. Alternative to which it is being compared

2. The criterion used (intelligibility / quality)

3. The signal level

4. The type of signal (speech, music, environmental sounds) the presence and type of noise and the signal to noise ratio

5. The frequency response shaping used

6. Hearing loss characteristics of the participants in the study

Compression affects the overall level of the output signal. The important factor in any comparison is how the volume control of each of the systems is adjusted.

Compression relative to linear amplification:

Most of the advantages and disadvantages are inevitable consequences of the changes in gain and changes in output level that accompanies compression. A compression aid with a low compression threshold and a linear aid have the same gain for a moderate input level. When a low level sound is input to both aids, the compression aid will have more gain, so its output will be more audible. On the downside, the compression aid will have a greater risk of feedback which could cause problems if there is enough leakage or sufficiently large vent. There is o theoretical basis for predicting how much compression is optimal. Hearing aid users will need to trade-off the increased loudness comfort and audibility against any extra amplification of background noise occurring in the gaps of speech and against any adverse change in the quality of speech or other signals.

It would be dicey to predict the effect of various compression rationale on intelligibility and comfort. It should be remembered that any compressor would increase the range of input sounds that fall within a person's comfort range without the use of volume control. Of course there is no theoretical basis for predicting how much compression would be optimal. Hearing aid users would need to trade off between the increased loudness comfort and audibility against extra amplification of background noise occurring in the gaps of the speech. Enu-

merated below are some of the pros and cons of various compression rationale used commonly in hearing aids under various circumstances.

1. Limiting - Also known as compression limiting which limits the maximum output of hearing aids. Compression limiting should be preferred to peak clipping, except for hearing aids intended for persons with profound hearing loss. Users in whom peak clipping cause distortion compression limiting is preferred. Peak clipping is advised only in individuals with profound sensorineural hearing loss. These patients are also ideal candidates for cochlear implants.

2. Typical input levels - If the user is prescribed a linear hearing aid he/she can adjust the volume control to get a comfortable loudness. For speech that is already at an optimal level in the absence of compression, slow acting compression does not affect the speech. Fast acting compressors can cause a decrease in the dynamic range of the speech but would not affect the overall intelligibility. Medium / low compression thresholds provide certain practical advantages over linear amplification. These include:

Listening comfort is increased in noisy places.

Need for a volume control is reduced.

Multichannel compression makes it easier to identify the manner of articulation of a consonant, but harder to identify the place of articulation.

3. Low level inputs - When the input sound is decreased (because the speaker has a soft voice) any form of compression threshold less than the original input level can provide intelligibility superior to that of the linear hearing aid. This is because of the fact that greater gain provided by the compression aid for low-level inputs thereby increasing audibility.

4. High-level inputs - If the input level is increased above the original level (above the compression threshold) any form of compression will increase listening comfort. Compression may also increase speech intelligibility because both excessively high presentation levels and peak clipping are harmful to intelligibility.

Use of compression in items 3 and 4 would considerably decrease the need of a manual volume control. This will actually be a boon for persons who have trouble manipulating volume control. If the compression is fast acting, the hearing aid would be able to decrease the rapid and large variations in level that can occur in some music. With linear amplification, these sounds would either be too soft or too loud. It should also be borne in mind that a hearing aid would not be able to tell the difference between a weak sound that is wanted and a weak sound that is unwanted. It would just turn up the gain whenever the sound remains weak long enough for the compressor to react. If this weak sound is a background noise then the compressor in the hearing aid would make this sound more noisier than a linear hearing aid. It is hence clear that compression has both beneficial and adverse effects.

Benefits of multichannel compression relative to single channel compression:

When compared to single channel compression, multichannel compression can increase intelligibility of speech because it increases its audibility. On the flip side the fast acting multichannel compression also decreases some of the essential differences between different phonemes. Like single channel compression multichannel compression

also flattens the envelope across the timeline. This is more effective because the flattening occurs independently in each of the several frequency regions. Since compressors give less amplification to intense signals than to weak signals, fast acting multichannel compressors also decrease the height of the spectral peaks and raise the floor of spectral valleys. Unlike single channel compression they partially flatten spectral shapes. Spectral peaks and valleys give speech sounds much of their identity. This spectral flattening makes it harder for the user to identify the place of articulation of consonants.

Fast acting single channel compression also has disadvantages relative to multichannel compression. Most obviously, gain variations produced by compressor affect all frequencies by the same amount and this gain is largely determined by the strongest frequency components present in the combined speech and noise at any instant. Weak components that may have little or no audibility can be attenuated just because a strong component is present in some other frequency. Another limitation is that for sloping hearing losses, the amount of compression has to make some compromise between that needed for frequencies where there is not much loss, and that needed for frequencies with more severe hearing loss. Single channel compression can cause signal and noise to have modulations in common and for these modulations to be consistent across frequency which makes it harder for aid wearer to distinguish speech from noise.

Slow versus fast compression:

Quality of sound may be maximized by slow acting compression while intelligibility may be maximized by fast acting compression. The importance of release time should be clearly understood as it gives the best sound quality and it varies from individual to individual. Studies reveal that individuals with low cognitive ability tend to obtain better speech intelligibility from slow acting compression than from fast acting ones.

Directional Microphones And Arrays

Directional microphones which function by sensing sound at two / more locations in space is an effective way to improve intelligibility in noisy environments. Directivity is commonly achieved in hearing aids with first order subtractive directional microphones in which the output of one omni-directional microphone is delayed and subtracted from the output of the other. The internal delay, relative to the physical spacing between the two microphone sound ports, largely determines the polar sensitivity pattern of these microphones.

There are only two proven ways of increasing intelligibility above that obtainable with appropriately adjusted conventional hearing aid delivering sound at a comfortable level. The first method is to move the hearing aid microphone closer to the source of sound. This step increases the level of direct sound compared to reverberant sound and background noise. Unfortunately this is not always practical. Another solution to this issue is to use some type of directional microphone. Directional microphones can be constructed from a single microphone with two entry ports or by combining the electrical outputs from two / more microphones. A microphone or group of microphones with more than one entry port is referred to as a directional microphone / microphone array / beam-forming array.

Technology involved in directional microphone:

Directional microphones widely used in hearing aids are first order subtractive directional microphones. This name is preferred because the output depends on a single subtraction of two signals.

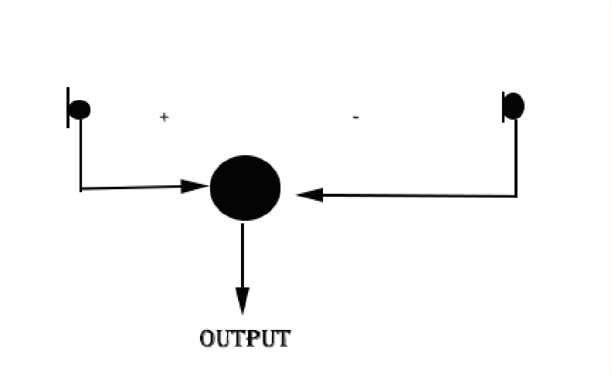


Image 1.63 showing a block diagram of a subtractive directional microphone comprising of either a single microphone with two ports or two separate microphones with one port each.

Port spacing, internal delay and polar pattern:

External delay - Sounds coming from directly in front or directly from the rear is calculated by dividing the port spacing by the speed of sound.

Internal delay - Is the delay that is integral to a low pass filter of the directional microphone or it can be an electronic delay.

The ratio of internal delay divided by the external delay will arrive at the delay ratio which in turn determines the shape of the sensitivity pattern which is also known as the polar directivity pattern. As the delay ration decreases from 1.0 to 0, the shape moves from a cardioid through a super-cardioid to a hyper-cardioid and then to a figure-8. Figure-8 is the extreme case and is a bi-directional pattern. The directivity index quantifies frontal sensitivity relative to average sensitivity and is fundamentally important to the understanding of how much benefit directional microphones can provide. The unidirectional index is similar but

less useful as it describes the sensitivity averaged across all frontal directions relative to the sensitivity averaged across all rearward directions.

The port spacing cannot be made too small, because the microphone itself becomes too noisy. It should be stressed that the microphone works by subtracting the pressure sensed at the two ports. The magnitude of this difference depends on the phase difference between the sound at the two port, and hence how large the port spacing is compared to the wavelength of the sound wave. Small port spacings therefore decrease the sensitivity of the microphone, but the internal noise generated by the microphone remains the same and so becomes increasingly apparent by comparison with the signal.

Head becomes more directional as frequency rises, the directivity pattern of directional hearing aids also varies with frequency when mounted on the head. The head combined with the design of hearing aid can have another effect. It is very common for the line joining the ports of the BTE aids to point upwards towards the front, which means that the direction of maximum sensitivity is also above the horizontal.

Directional microphones can be used only in hearing aids large enough to accept the necessary port spacing. At present they are mostly used in BTE and ITE hearing aids. They are not likely to be effective in CIC aids because there is not much room to achieve the necessary port spacing and partly because diffraction by the pinna creates a complex sound field near the faceplate of the hearing aid.

Frequency response:

Since wavelength progressively lengthens as frequency decreases, a fixed port spacing represents a smaller and smaller fraction of a wavelength as frequency decreases. Consequently, the sound pressure at the two ports at any instant becomes more similar, so the difference between the front and back port signals decreases. A subtractive directional microphone therefore has a low-frequency cut of 6 dB per octave in the gain frequency response. An electronic filter can be used to boost the low frequency gain and so compensate for this. Such a filter could also boost the internal microphone noise making it more annoying to the user. To avoid this excessive noise, it is common for hearing aids to compensate only partially for the low frequency cut.

Compensation actually is most important for patients with more than 40 dB hearing loss in the low frequencies, who rely on amplified low frequency sound, and is least important for high frequency. Microphone sensitivity can be maximized, and internal noise minimized by using a large port spacing. But if made too large the frequency response for frontal sounds could be adversely affected.

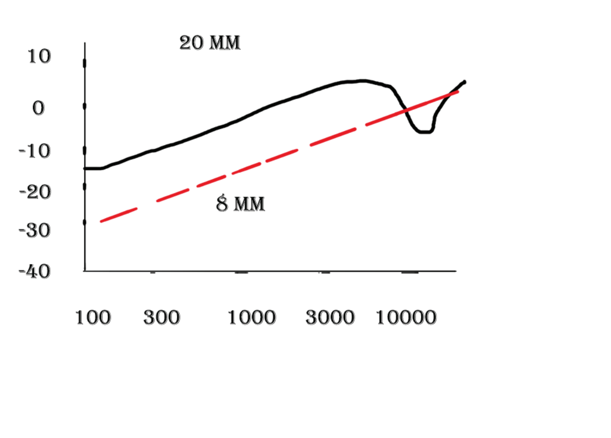


Image 1.64 shows the frequency response for microphones with a 20 mm and 8 mm port spacing. The minimum in the high frequency response occurs at 9 kHz for the larger port spacing, but at a

frequency far above the bandwidth of the hearing aid for the smaller port spacing. It is also evident that the 8 mm port spacing has a lower sensitivity over most of the frequency range as expected.

Acoustic versus electronic subtraction:

Hearing aids with directional microphones nearly always enable an omni-directional response to be selected either by incorporating both a directional microphone and an omni-directional microphone or by using two omni-directional microphones the outputs of which are combined electronically.

Even though manufacturers select pairs of well matched microphones, the responses of the microphones change after manufacture due to combined effects of age, humidity, temperature, vibration and accumulation of debris in the microphone ports. The resulting mismatch in gain-frequency response causes a progressive deterioration in their matching and hence directivity. For example a sensitivity mismatch as small as 1 dB or a phase mismatch as small as 5 degrees can decrease the directivity index by one third at 1 kHz, with progressively much worse effects as frequency decreases below 1 kHz. This mismatch problem is minimized electronically in many hearing aids by the hearing aid continuously comparing the long term average output of the two microphones.

Commonly hearing aids can automatically select a directional response in some situations and an omni-directional response in others. If there is a dominant speaker in one direction, and a variety of sounds from other directions the hearing aid would assume that the wearer would like to focus attention to that sound and would adjust the settings automatically.

Hearing aids can simultaneously process both a directional and omni-directional signal and choose the one that has the higher apparent SNR. The types of information that hearing aids take into account to determine that a directional response is most appropriate are:

1. The overall sound level is high enough to indicate that voice levels are raised.

2. The background sound level, measured during apparent gaps in the main signal, is greater than some amount, typically around 60 dB SPL, suggesting that there is noise in the environment and also making it unlikely that the internal noise of a microphone will be inaudible.

3. The output of the directional microphone has deeper envelope fluctuations especially at the rates typical of speech (4-20 Hz) than the output of the omni-directional microphone suggesting that there is a speaker somewhere frontal of the hearing aid wearer.

There could be some situations where the aid user wishes to hear a talker behind him. In such situations, an omni-directional pattern will be better than a directional pattern. Directional pattern microphone pointing backward would be even better. This is also known as reverse cardioid.

Additive directional arrays:

This works on a principle different from subtractive arrays. Instead of reducing sensitivity for sounds from all directions, but reducing it least from the front the additive array produces the maximum possible sensitivity for sounds coming from the front and less sensitivity for all other directions.

This is effective only for high frequency sounds.

Those for which the length of the array is greater than or comparable to a quarter wavelength.

End fire arrays have microphones arranged in a line that goes through the most sensitive direction, whereas broadside arrays are in a line that is perpendicular to the most sensitive direction.

Additive arrays combine microphone outputs by addition, whereas subtractive arrays combine by subtraction.

Fixed arrays have the same polar sensitivity in all situations, whereas adaptive arrays have a sensitivity pattern that changes with the direction of the surrounding sources and optionally reverberation characteristics.

Second order arrays

Subtracting the outputs of two microphones can produce a directional pattern. Similarly subtracting two signals each of which is the output of the first order subtractive directional microphone can produce super-directional pattern. The act of fitting a wide bandwidth second order subtractive array within even a large BTE would need a port spacing so small that the internal microphone noise would be really excessive. The second order processing can be incorporated into BTE by restricting the second order processing to high frequencies. Even though in these hearing aids the port spacing is rather small, the internal noise and microphone mismatch is not a problem.

Adaptive weight arrays

The most advanced and effective arrays are known to vary the gain and phase of each microphone output by different amounts at different frequencies, before adding all these filtered microphone outputs to produce the output signal. These adaptive filtering is used in commercial hearing aids and cochlear implants. When the source of noise is only one, no reverberation and a very poor SNR the adaptive filter can change its characteristics so that the directivity pattern of the array has a perfect null in the direction of the noise while keeping the normal sensitivity in the target direction. Under favorable conditions, the SNR can be improved by as much as 30 dB.

Reverberation greatly decreases the effectiveness of adaptive arrays. Unless the speaker is very close, reverberation will cause significant speech energy to arrive in all directions. Hence the noise reference signal will contain speech as well as noise. This mixture makes it difficult for the filter to adapt, thus reducing the effectiveness of noise canceling. The subtracter will also remove some of the speech as well as the noise thus affecting the quality of hearing spoken words. A voice activity detector will help in resolving this problem. This is based on the pulsatile nature and overall level of speech signals. This voice detector would stop the adaptive filter from changing its response whenever speech is believed to be present so that noise alone determines the filter weights.

Bilateral directivity

Often people with bilateral hearing loss tend to wear hearing aids in both ears. When the target speech arrives from the front and noise arrives equally from all directions, most of the benefit can be provided by either hearing aid having a directional response.

Directional benefit

Benefit provided by a directional microphone depends on the directivity of the hearing aid, the

reverberation characteristics of the listening situation, the distance of the talker and noise sources. The directivity of the hearing aid depends on the openness of the fitting, the gain of the hearing aid, and its position within or behind the ear.

Advanced Signal Processing

Several advanced signal processing techniques are used in modern hearing aids. The following are some of these techniques:

Adaptive noise reduction technology

This involves improving speech intelligibility by noise removal. This is actually a difficult problem to surmount because our understanding of this problem is rather limited and hence have been rather unsuccessful in finding a solution. Noise of course can be reduced in such a manner that it improves listening comfort, and reduces listening effort. There are many techniques that can be used to accomplish this. Various synonyms are being used to denote this technique including noise suppression, fine scale noise canceling, single microphone noise reduction and digital noise reduction.

Basic aim of adaptive noise reduction is to provide less amplification to noise than to speech. This is achieved by identifying segments where noise is particularly intense relative to speech, and applying less amplification to these segments than to other segments where the SNR is better. If this is achieved, it is likely that the hearing aid wearer will find the noise less troublesome. If the noise is intense the device will reduce it thereby minimizing the likely-hood to speech frequency masking. For noise reduction to become possible, the hearing aid has to detect speech, eliminate the speech and noise levels.

Detecting speech and noise

Speech sounds are created by successive opening and restrictions in the vocal tract causing random turbulence of airflow across the restrictions. This variation / modulation of the amplitude of speech is the primary characteristic used to detect the presence of speech. The amplitude of speech thus increases and decreases with a frequency varying across the 3-6 Hz range. When speech power is the maximum, the power is likely to be large simultaneously at many frequencies so the same 3-6 Hz fluctuations should be detectable in more than one of the hearing aid channels. This is known as co-modulation which provides a second clue that speech is present. The third clue arises from the fine texture of speech. Every time the vocal cords open, a burst of power across a wide frequency range occurs and these bursts repeat at the pitch rate of the voice (around 100 Hz for male up to around 400 Hz in a child). The presence of these bursts, at typical pitch rates, synchronized across multiple frequency channels within the hearing aid provides further confirmation that speech is present.

Estimating speech & Noise levels

Modulations are used to detect speech can also be used to estimate the signal to noise ration within each channel.

Figure 1.65 shows the envelope of a speech signal in the presence of noise. The eye can quickly discern that there is something present with a relatively constant level around 55 dB SPL during gaps between each word. This is actually the steady background noise, whereas the peak levels of the envelope represent the maxima and minima of the envelope and the difference between them is the modulation depth of the signal. These operations are carried out separately within each hearing aid channel, so the modulation depth in each frequency range is known.

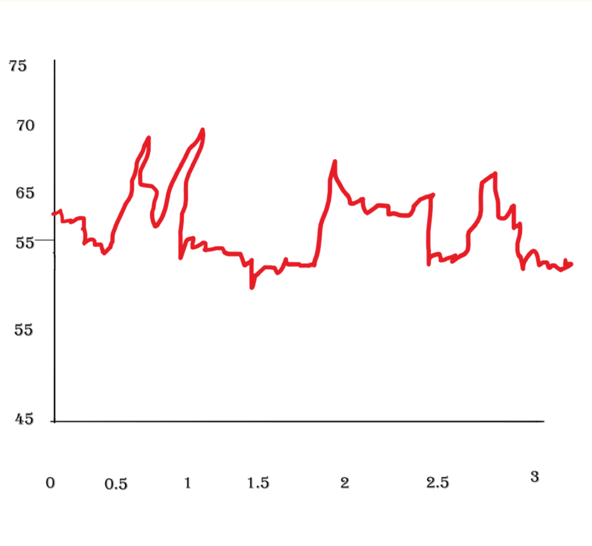


Image 1.65 showing envelope of the sentence, “The yellow flower has a big bud” in the presence of background noise with a steady level of 55 dB SPL.

If the maxima and minima are averaged across several seconds, the long-term SNR can be estimated. If the near-instantaneous maxima are used, then the short-term SNR, which varies rapidly can be estimated.

The value of SNR is smaller than that of the depth of modulation (typically about 10 dB), because the SNR difference between the average speech level and average noise level. Modulation depth is the difference between the peaks of the envelope and the average level of the noise. This modulation depth approach to estimate SNR works well when the signal is from a single speaker and the noise is a continuous babble, or other noise with few fluctuations in level at the syllable / word rate. Hearing aid is likely to suffer as far as SNR is concerned when the needed signal has little fluctuations (some types of music), and the noise has marked fluctuations (from a single nearby speaker).

Information about the presence of speech and estimated SNR is used by some hearing aids to automatically select the most appropriate microphone configuration (directional versus omni-directional, or the most appropriate polar pattern), as well as to control adaptive noise reduction.

Gain reduction algorithms

There are a number of ways available to perform adaptive noise reduction. Majority of systems use either Wiener filtering or Spectral subtraction.

Wiener filter is a filter whose gain at each frequency depends in a particular way on the SNR at that frequency. Specifically, the gain equals the signal power divided by the sum of the signal power plus noise power.

$$W(f) - \frac{s(f)}{s(f)+n(f)}$$

s(f) - power spectrum of the signal

n(f) - power spectrum of the noise

It has been demonstrated that Wiener Filter makes the waveform at the filter output as similar as possible to the signal (without noise) at the input. The essential characteristic of the Wiener filter is that gain is reduced as the SNR deteriorates.

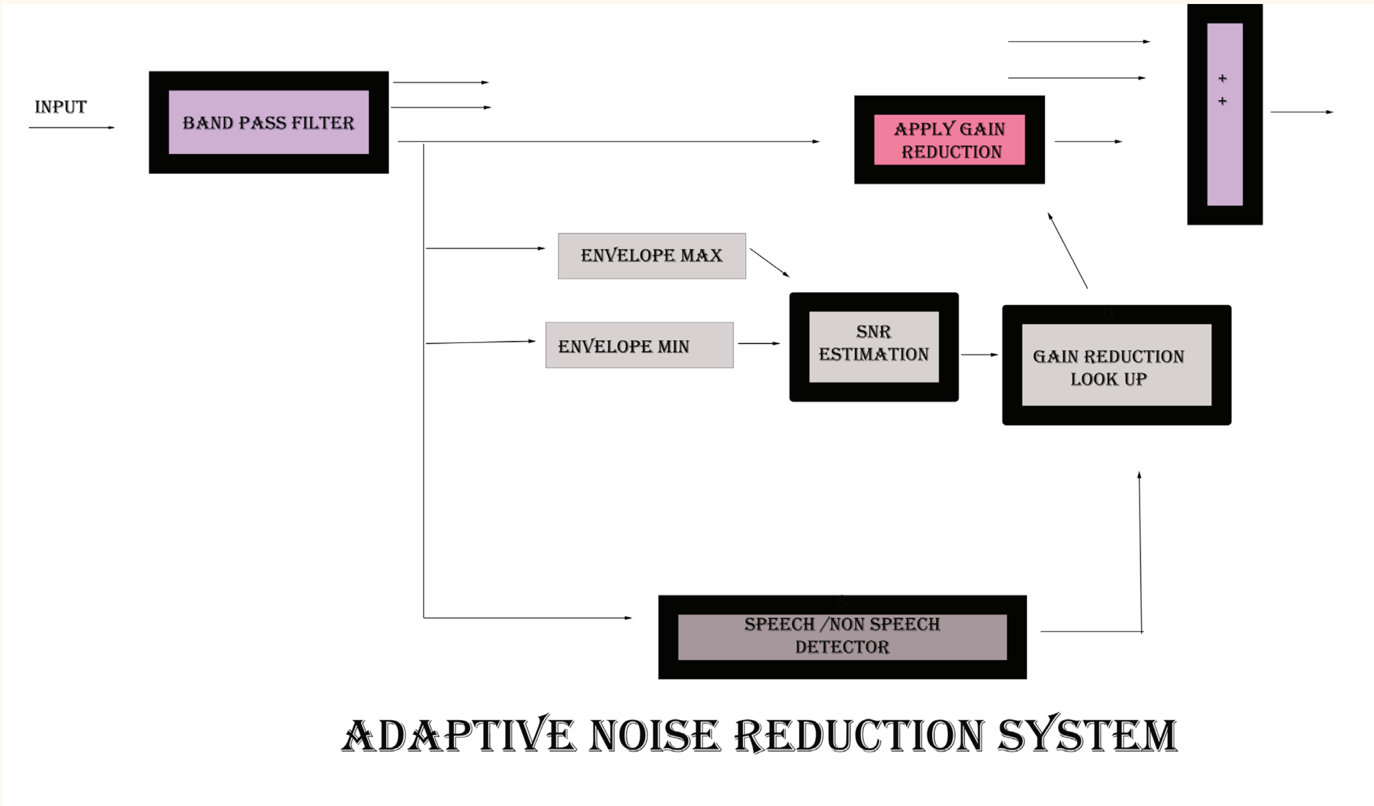


Image 1.66 showing adaptive noise reduction system

The signal processing needed to accomplish adaptive noise reduction, using modulation depth to estimate SNR, is not complex and this is amply illustrated in Image 1.66. It should be remembered that there is not much research available to guide exactly how much gain should be reduced at any SNR. The optimal gain changes are not insignificant. It is important that gain reduction should never be of sufficient degree that it decreases the amount of speech that is audible, as it would decrease the intelligibility of speech. In principle, the gain reduction should be marked at the highest input levels, and least marked for the greatest hearing losses. This can be achieved by ensuring that attenuation should be never greater than the amount that causes the noise level to be decreased to hearing threshold.

Spectral filtering

This is an alternative to Wiener Filtering. The magnitude of noise spectrum is subtracted from the magnitude of the speech plus noise spectrum. If both these magnitudes are known exactly, then the difference will be the magnitude of the speech spectrum alone. The problem with this system is determining the spectrum of the noise, because the microphone picks up the speech and noise combined. One obvious solution to this problem would be estimating the noise spectrum present currently by averaging the noise spectrum that was present during some preceding moments in time as in the case with Wiener filtering. This spectral subtraction system needs a speech/non-speech detector. Only the magnitudes of the speech plus noise signal are corrected by the processing.

Although Wiener filtering and Spectral subtraction work on different principles, they have similar effects on a noisy signal. Both decrease the gain most at those frequencies where the SNR is worst, and leave the signal unaltered when there is little noise present.

Adaptive noise reduction systems can be designed to vary the gain reduction applied every few milliseconds or take many sounds before they respond in any way to noise, and then vary the gain gradually over a duration of several seconds.

Onset time is defined as the time from when noise commences to when the gain has reduced to within 3 dB of final value. The onset time can vary from a couple of seconds to more than 30 seconds.

Offset time is defined as the time from when noise ceases to when the gain has been restored to within 3 dB of the value it has in quiet. This offset time can vary from 5 ms to several seconds.

Spectral subtraction is inherently very fast acting, as the subtraction occurs separately for each brief segment of speech analyzed. Analysis frames in hearing aids are typically 4-8 ms long. Wiener Filtering can also be implemented in such a way that gain in each frequency region varies every few milliseconds or the rate of gain variation can intentionally be slowed down so that it takes several seconds for the gain to change significantly. This slow approach is intended to react to changes in the listening environment.

The main advantage of fast-acting noise reduction is that noise between words and syllables in the speech is reduced, not just noise in frequency regions where the noise dominates. The main disadvantage of fast-acting noise reduction is that the rapid changes in gain can distort speech quality. Fast-acting noise reduction hence has the greatest potential to improve speech comfort, its intelligibility. It of course runs the greatest risk of producing processing artifacts, especially if the speech detector mistakes lower level speech components for noise.

Benefits of adaptive noise reduction:

Adaptive noise reduction systems can be expected to improve the overall signal to noise ration when the levels of the signal and noise are measured objectively at the output of the hearing aid.

Image 1.67 shows an example where the SNR at the output will be much greater than that of the input. This improved SNR does not generally cause any increase in intelligibility of speech.

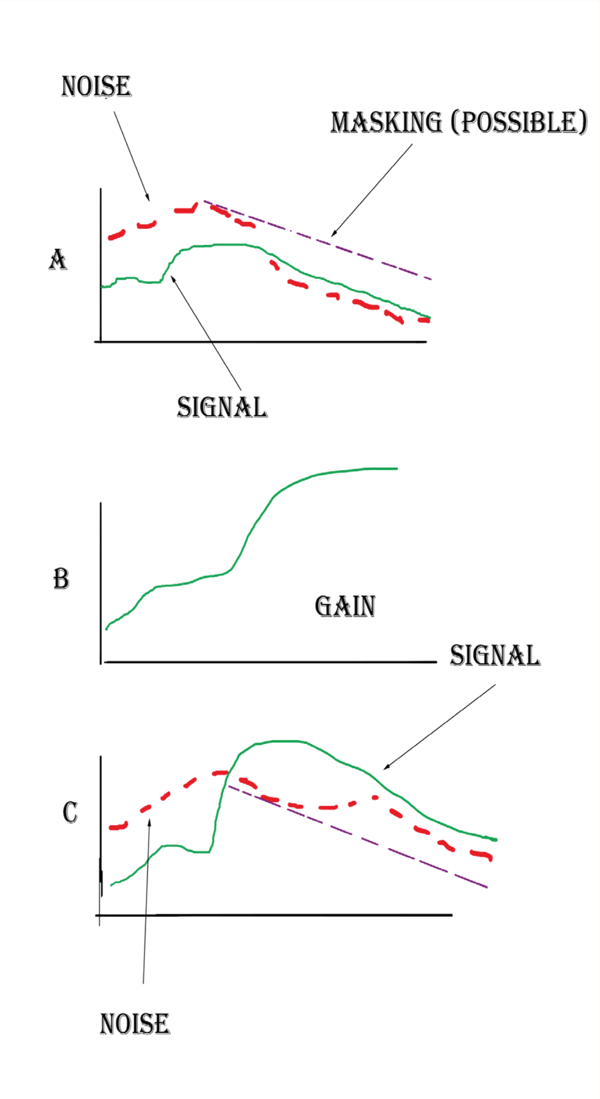


Image 1.67 showing SNR at the output much higher than that of input.

A - Input

B - Gain

C - Output

If the hearing aid has a single microphone port, then when a noise and signal occur at the same time and at the same frequency there is no known way by which they can be separated. Improving intelligibility in noisy environment with only a single input relies on the signal and noise having components sufficiently different in frequency or in time to be separable by signal processing. This could be achieved, especially in persons with severe and profound hearing loss as they have the most reduced frequency and temporal discrimination abilities. A solution to this problem has been elusive.

If noise is restricted to a narrow frequency region, adaptive noise reduction can lead to a substantial increase in intelligibility, as attenuation of this narrow frequency region can decrease the masking caused by noise. Despite the general lack of speech intelligibility benefits from adaptive noise reduction and processing is almost always preferred as it is comfortable, provides better ease of listening, better quality and overall performance.

Despite the fact that there is no expectation of improved speech intelligibility, adaptive noise reduction should routinely be enabled whenever the hearing aid senses that significant levels of background noise is present or is likely to be present. This could enable the user to communicate for longer duration in noisy situations.

Impulse noise reduction

Speech sounds are produced by the vocal tract. There are of course limits to how rapidly the vocal tract could change and hence how rapidly the instantaneous pressure of a speech waveform can change. A smart hearing aid could recognize when the pressure is changing too rapidly for the signal to be speech, and deliberately not reproduce the rapid rise and fall of such sounds. This im-

pulsive smoothing of sound / transient loudness reduction would hence reduce the loudness, and hence annoyance of the impulse sound without completely removing it, while having little or no effect on any speech sound that occurs at the same time. This technology gives increased loudness comfort, with no change in intelligibility.

Feedback reduction

The causes for feedback has already been discussed elaborately in previous chapters. Several electronic methods are available for mitigating the effects of feedback. All of these methods could help, but none of them can totally remove feedback oscillations completely.

Feedback reduction by gain-frequency response control

Feedback oscillation of a specific frequency occurs when the gain from the microphone inlet to the ear canal is greater than the attenuation from the ear canal back to the microphone at that frequency. Furthermore, at this frequency, the phase shift around the entire loop should be close to an integral number of periods. One way to avoid feedback oscillations is to decrease the gain at all frequencies where these conditions are met. This can be achieved by several ways. The simplest way is to turn the volume control down below the point required by the patient. This of course could be unsatisfactory, as it would give the patient inadequate loudness, audibility, and intelligibility. A better alternative would be to decrease the gain at only those frequencies where feedback oscillation is a possibility. This is most likely to be at or near the peaks of the gain-frequency response curve, so anything that decreases the gain at these peaks without reducing the gain elsewhere is likely to be beneficial. Acoustic damping in the sound tube meets this criterion very well. It may not be always possible to damp the particular peaks causing the feedback oscillation without excessively decreasing the gain in the frequency region around some other resonances.

Multichannel hearing aids provide a more reliable way to decrease gain in only one frequency region. This degree of control over the gain-frequency response is extremely coarse, however, unless the hearing aid has many parallel channels. If there are only a few channels, gain may be decreased over an unnecessarily wide frequency range resulting in inadequate audibility.

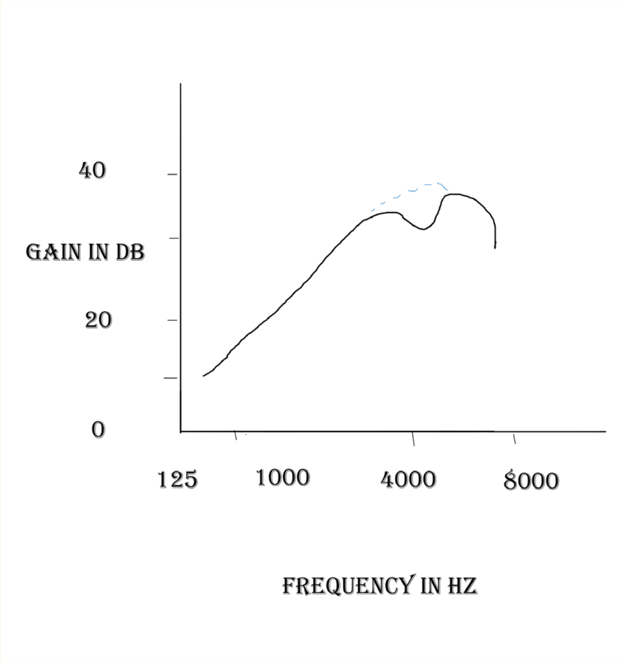


Image 1.68 showing the gain-frequency response of a four channel hearing aid where feedback oscillation has been avoided by decreasing the gain of the band from 2 kHz (solid line) from the original response (dashed line).

Commonly feedback occurs only when the volume control is increased above the aid wearer's usual setting or at low input levels because of the effect of wide dynamic range compression on gain. In such cases it is necessary to decrease the gain in a narrow frequency region only under these specific conditions. Hearing aids can hence avoid feedback oscillation by limiting the maximum gain that can be achieved in each frequency region. the limit depends on the tightness of the fit of the earmold or ear shell. When feedback is not a problem, the full desired gain-frequency response is provided to the hearing aid wearer. When the overall gain is increased either manually or automatically by compression then gain in frequency regions likely to cause oscillations can be held down to a safe value.

Determination of safe value:

This value is usually determined at the time of fitting. This is calculated by the clinician selecting the maximum gain that just avoids oscillation, or by increasing the compression threshold (and hence the gain for low input levels) until oscillation ceases.

This value can also be calculated by performing an in-situ feedback test, in which the fitting system automatically raises the gain in each channel until it detects oscillation occurring.

The hearing aid can also be tuned to reduce the gain in a channel whenever it detects oscillation occurring in that channel, and allowing full prescribed amplification when oscillations are not present.

Digital filters can provide faster control of gain-frequency response shape. Once the frequencies that can cause feedback oscillation are identified, narrow notches can be placed in the gain-frequency response around each of these frequencies. This technique is commonly used in all Public Address systems. The frequencies at which feedback occurs do not remain fixed over time.

Reasons for using electronic feedback control

Electronic feedback control could be useful in the following circumstances.

When more gain is needed. This is particularly the case in persons with severe to profound hearing loss or persons would like a smaller hearing aid style than could otherwise be provided without feedback.

When a more open earmold or earshell is needed. This is useful for persons with mild hearing loss at low frequencies and severe loss at higher frequencies.

Gain reduction if implemented represents a loss to the patient. The degree of loss is minimized if the gain reduction is adaptive - that is if it occurs only when oscillation is detected. Many hearing aids continually monitor their output to detect feedback oscillation, measure the oscillation frequency, and automatically adjust the gain-frequency response to prevent oscillation from continuing. This type of automatic / adaptive gain reduction systems are commonly known as search and destroy feedback control.

Feedback reduction by phase control

Phase variation could help in prevention of feedback oscillation. In principle, the phase can be intentionally manipulated to reduce the likelihood of feedback oscillation. The aim of phase control is to ensure that at any frequency where the gain is large

enough to cause oscillations, the phase response around the loop causes the feedback to be negative rather than positive.

Previously hearing aid designers used rudimentary control of phase in analog devices. Reversing the connections to the earphone (if possible) adds 180 degrees to the phase response, which in 50% of the time will allow a greater gain to be achieved without oscillation at least for some settings of the tone controls.

Feedback reduction by feedback path cancellation

The most effective, widely used technique in digital hearing aids is feedback path cancellation. This method intentionally creates a second feedback path, completely internal to the hearing aid. This internal path has the right gain and phase response to cancel the external leak path. If at any frequency, the two feedback paths leak back the same amount of signal, and if these two signals have the same phase, they will sum to zero, and there is no net feedback. Without any feedback, there can be no oscillation.

This appears to be a perfect solution, but like any other solutions, it can increase the maximum stable gain only to a certain extent. This increase in maximum stable gain enabled by the feedback cancellation algorithm is referred to as added stable gain (ASG). The more closely the internal path matches the external leakage path, the greater the ASG.

In order to achieve this high added stable gain (ASG) in daily life, changes in the characteristics of the leakage path over time should be allowed. There are two possible ways this can be achieved:

The first method which is no longer used, a test signal is injected either with or without the amplifier chain being broken in order to measure the characteristics of the external feedback path. This protocol allowed the gain to be increased by approximately 10 dB before feedback commenced. The basic problem with this methodology is that the test signal was audible to the wearer, expect in case of profound hearing loss. This method is included here just for the sake of completion.

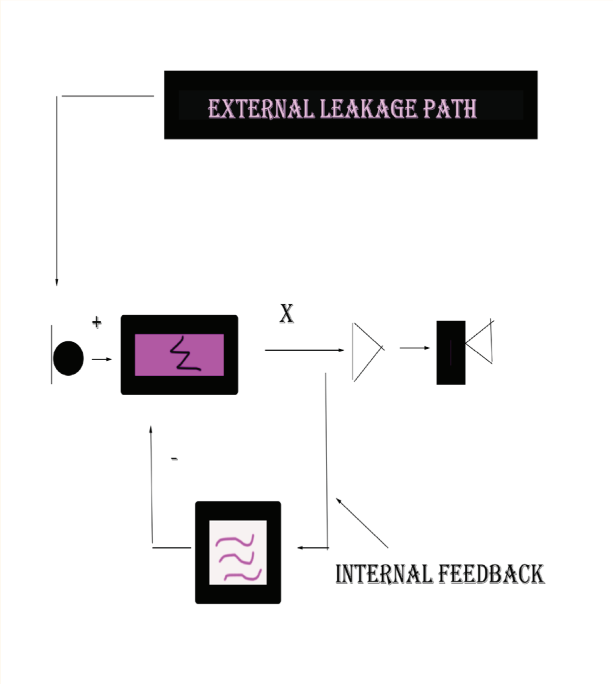


Image 1.69 showing internal feedback added to cancel the effects of the external, unintentional leakage path. The filter adapts so that it minimizes the evidence of feedback at point x.

The currently used method shown in image 1.69 automatically adapts in such a way as to minimize any signal that continues at a single frequency for more than a certain amount of time, such as feedback oscillation, or low level ringing caused by sub-oscillatory feedback. The hearing aid automatically measures the external leakage path during fitting, and hence initializes the internal canceling

path. Real time measurement of the output signal is used to continuously fine-tune the internal path. This real-time measurement is necessary if the algorithm is to cope with any of the events that normally cause a stable hearing aid to suddenly begin oscillating or to cause sub-oscillatory ringing. These could include movement of the earmold in the ear canal, presence of a reflecting surface near the ear, or low levels of background sound that cause the hearing aid gain to increase.

A major advantage of feedback path cancellation over gain-frequency response control is that accurate feedback path cancellation does not cause any decrease in gain. If a hearing aid is close to oscillating, the positive feedback provided by the external leakage path increases gain. The internal negative feedback path causes a corresponding decrease in gain and the combination of the two leaves the hearing aid with the same gain at all frequencies in the absence of feedback of any sort. A second advantage is that the cancellation can work even before the hearing aid continuously oscillates. Measurement of the output signal can detect and remove sub-oscillatory feedback or ringing.

There are some disadvantages to this feedback cancellation circuitry. While the filter is adapting (this happens whenever the leakage path changes) the filter could distort speech quality during the second or so that it takes the filter to adapt. Another major disadvantage of this system is that the feedback canceler will also cancel other sustained periodic signals, (somebody whistling or the sound of many musical instruments). Many enhancements have been added to high end hearing aids to minimize the distortions the feedback cancellation can cause to musical sounds. These include:

1. Feedback canceling algorithms can be disabled, or the rate of adaptation slowed down. This can occur when the program selector in the hearing aid is used to select music program.

2. The forward path of hearing aids can incorporate a small frequency shift, or a rapidly varying phase response, both of which help the hearing aid in distinguishing between internal feedback and external sounds.

3. The hearing aid can compare the sound levels reaching each of the two microphone ports or even more effectively the two hearing aids on the opposite sides of the head. External sounds will result in the level at the microphones being more similar than is the case for oscillation.

Feedback cancellation requires more calculations than gain reduction feedback control. More the number of calculations needed more will be the battery usage. Battery life is one important constraint when these technologies are used.

Benefits of feedback path cancellation are substantial. Hearing aids are most likely to oscillate in quite environments, where compression causes the highest gains. Unfortunately, quite environments are precisely the environments where hearing aid user can least afford to reduce the hearing aid gain. Hence additional gain enabled by feedback cancellation should translate into a real intelligibility advantage in quiet environments.

Close proximity of a telephone headset normally requires a significant gain reduction. Feedback cancellation reduces the amount by which gain has to be reduced and hence improves speech intelligibility over the phone.

Hearing aid designers usually prioritize achieving the greatest added stable gain as possible, and risk the possibility of signal distortion, or a very

slow response when the leakage path changes. Some manufacturers emphasize a fast response or minimal possibility of distortion and hence do not achieve much added stable gain values. Room reverberation, with its very long delays provides a limit on the added gain response achievable. Each time the user moves within the room, even slightly, the feedback path changes and so too must the internal filter if it is to accurately mimic and cancel the external leakage signal. Added stable gain value that is achievable without audible artifacts is affected by the presence of other adaptive signal processing in the hearing aid.

Feedback reduction by frequency shifting

Feedback oscillation occurs if a sound gets larger every time it goes around the feedback loop. One scenario that is quite possible is if a sound comes out of the amplifier at a different frequency than that went in. In this case the signal leaking back to the microphone would be at a different frequency from the original input, the two sounds could not remain continuously in phase with each other and so could not build up in amplitude as effectively as possible. Consequently, the likelihood of feedback would be considerably lessened.

Feedback reduction systems in combination

It is common for hearing aids to include more than one method of feedback control. It is in fact desirable. A precise, and slow acting feedback cancellation system can provide a large added stable gain. It could take 10 seconds to compute the new internal filter characteristics needed when the when the hearing aid user puts on a hat. During this period a fast acting adaptive system can detect the oscillation and reduce the gain at the appropriate frequency in less than a second. The aid wearer thus could experience little whistling, and the gain reduction is no longer needed once the slow acting system completes its estimation of the altered external feedback path. Feedback cancellation algorithms can operate with more aggressive settings if the processing also incorporates frequency lowering.

One of the major advantages of digital hearing aids has been the availability of effective methods of feedback reduction, especially feedback path cancellation.

Frequency lowering

Most hearing impaired individuals have a greater loss for high-frequency sounds than for low-frequency sounds. In some patients the high frequency loss is so great that they cannot extract any useful information from the high-frequency parts of speech. Worse still is that in some of these patients excessive amplification of the high frequency parts of speech can decrease their ability to recover useful information from the low and mid frequency parts of the speech signal. In these patients any chance of accessing the information that exists only in the high-frequency parts of speech, the information must be moved down to some other frequency region where the person is more able to analyze sounds. This is the basis of frequency lowering in haring aids.

Rules for lowering frequencies by hearing aid

A simple technique would be to reduce all information in frequency by some constant number of Hz. This is referred to as frequency transposition. An example of this concept would be all information could be presented 2 kHz lower than it originally was. The problem with this approach is

that input power from 0-2 kHz cannot be lowered this way and hence need to remain at the original range. For sounds with significant energy below and above 2 kHz, such as voiced fricatives, the result may be confusing and ambiguous.

Input sound with energy over the whole range, the spectral shape of the output will be a complex mixture of the different input frequency ranges, here important features like formants, originating in one frequency band may be obscured by speech components originating from other frequency band. Many individuals with severe high frequency hearing loss consider that transposition improves speech clarity in them.

One method that can be followed to minimize the problem of transposed sound sharing the same frequency range with unmodified sound is to apply transposition only when the input spectrum is dominated by high frequency components. This process is termed as conditional frequency transposition / dynamic speech encoding. Transposed energy will then be available for sound for which it is most needed, and will not produce adverse effects when low frequency sounds are present. This method can increase the intelligibility of stops, fricatives, and affricates without degrading the intelligibility of nasals and semi-vowels.

Frequency compression

This is an alternative to transposition and has the advantage of no overlap in output spectra. This method is known as frequency compression. When the output frequency is a constant fraction of the input frequency then it is termed linear frequency compression.

Frequency compression produces undesirable effects if applied to the entire frequency range. Although linear frequency compression preserves the correct ratios between frequencies, which helps to preserve the identity of vowel sounds and the voice like quality of speech. After transposition is applied, young children may sound like female adults, and female adults could sound like males.

Some of the frequency lowering techniques used in hearing aid

Modulation

This simple technique used for frequency lowering in hearing aids is distorting sound such as with pronounced peak clipping. More sophisticated modulation method involves selecting a frequency range by filtering, multiplying a pure tone by the filtered signal, and selecting by filtering one of the sidebands created by the modulation. Multiplying the 4 to 8 kHz pure tone moves the range down by 4 kHz to the range between 0-4 kHz. Modulation is suitable for linear transposition.

Slow playback

If a signal is played back at a slower rate than the one at which it was recorded, all frequencies are reduced by the same proportion. Slow playback is easily accomplished with digital processing. Practical problem with this type is that the processed signal being slower, would get further and further behind reality unless some correction is applied. Obvious solution to this problem could be deletion of sufficient sections of the signal such that the original and slowed down signal have the same duration. Ideally, complete voice pitch periods are deleted so that waveform remains continuous and artifact free.

Speech vocoder

In this process speech is filtered into a bank of adjacent narrow bands, and the level within each band is detected. Speech can be re-synthesized using these levels to modulate the level of narrow bands of noise, or pure tones at the frequency of each original narrow-band filter. A frequency lowering speech vocoder is constructed by making the frequency of the narrow bands of noise or pure tones lower than the center of the bands from which the signals modulating them are derived.

Adaptation

Frequency lowering scheme could make speech sound different. It takes sometime for the user to become accustomed to this effect and benefit from this form of sound processing. It should be pointed out that children are more adaptable than adults at quickly making use of the altered cues that frequency lowering provides. Greater listening experience will be needed for identification of sounds in continuous discourse than for simple differentiation of different consonants. Training could facilitate and accelerate adaptation to the new cues.

Parameters for frequency lowering

One good strategy is to lower all frequencies for which hearing thresholds exceed some predetermined value. A good strategy would be to lower frequencies that occur deep within the dead region of the cochlea to a band around the lower edge of the dead region. Frequency lowering increases the ability to detect high-frequency consonants especially /s/, and hence help in differentiation of plural from singular nouns.

Speech cue enhancement

Algorithms are available to modify speech in order to make it more intelligible for people suffering from sensorineural hearing loss. Any acoustic feature of a speech sound that helps in identifying it can be detected and exaggerated to make recognition of that feature a bit easier.

Enhancement of spectral shape

Attempts have been made to detect the prominent spectral peaks of speech sound (the formants) and amplify them preferentially. This feature is known as spectral contrast enhancement / spectral sharpening. This causes the formant structure to become sharper and the locations of formants on a spectrogram become better defined. Experiments have demonstrated that emphasizing small spectral peaks by surrounding them with deeper spectral valleys help people with hearing loss in detecting the original peaks. Another form of spectral enhancement which is rather extreme is sinusoidal modeling. Here the most dominant spectral peaks are replaced by pure tones with the appropriate frequency, amplitude and phase. This method is rather effective in reducing background noise. This can also be viewed as a form of speech simplification.

Enhancement of consonant-to-vowel ratio

The ratio of consonant level to vowel level is referred to as the consonant-to-vowel ratio. This ratio is negative in unprocessed speech. This ratio can be increased by preferential amplification of the consonants and not the vowels. When this value is increased, speech intelligibility does not generally improve, indicating that the absolute level of the consonant is important and not their level relative to the vowels.

Transient enhancement

Intensity enhancement has been linked to the rate of change of intensity. Many consonants with a low level relative to adjoining vowels have rapid intensity changes often with accompanying spectral changes that must be perceived for the consonants to be correctly identified. There is a potential to increase intelligibility if the rapid variations in intensity can be made more prominent.

Increased prominence can be achieved by a circuit that automatically increases its gain whenever the intensity of the input signal is changing rapidly. The resulting speech is perceived as though all plosives have been articulated with great emphasis. One potential problem is that consonants with a somewhat rapid onset followed by a steady level can have their onset intensity enhanced to such an extent they sound like an affricate with a similar spectrum. The word ship sounds like chip. Transient enhancement may be beneficial for hearing aid wearers with auditory neuropathy as they have significantly reduced temporal resolution.

Enhancement of duration

Another feature that has been modified is the length of the vowels. Vowels preceding a voiced consonant tend to be longer than vowels that precede an unvoiced consonant. These differences in the preceding vowel length are one of the cues used by normal hearers to distinguish voiced from unvoiced consonants. Since perception of duration is little affected by hearing loss, vowel length is a particularly important cue for people with hearing impairment for whom spectral differences are less clear. The decision to lengthen or shorten the vowel must be made before the vowel ends and therefore before the final consonant has even reached the hearing aid.

Speech simplification

Persons with profound hearing loss are unable to perceive many of the complex cues in a speech signal, particularly when noise is present. They will understand more if less information is presented. Speech simplification just plays this role. This is the basic concept behind speech pattern processing. At one extreme, the speech signal is replaced by a single pure tone pulsing on and off in time with the speech. The pure tone has a frequency equal to the fundamental frequency of the speech. Other features that have been extracted from the speech and presented in a simple manner include the amplitude of the speech envelope and the presence of voiceless excitation. Identification of speech is better with these additional features than with fundamental frequency alone. Presentation of a simplified speech code also helps profoundly hearing impaired persons to control the fundamental frequency of their own voice.

Enhancement by re-synthesis

Another example of using the special features of speech is a hearing aid that recognized speech and then re-synthesized it in a clear, well articulated, and noise-free way. There could be problems with this system as automatic speech recognizers do not perform well in noisy places and have trouble with unusual accents.

Other signal processing schemes

Digital signal processing is available in a range of other schemes. Some of them are already being used in hearing aids.

De-reverberation and echo reduction

Reverberation is highly detrimental to speech intelligibility, for normal hearing and hearing impaired persons alike. The reverberation from each phoneme masks some of the power in the following phonemes, especially when the following phonemes are lower in level than the earlier phoneme. Processing to reduce reverberation, even when it is overlapped with other sounds is technically possible. This needs the processor to know the electro-acoustic characteristics of all the signal paths from the talker to the listener. Reverberation does not overlap a succeeding sound and can be recognized by its characteristic gradual drop off in intensity with time. Once recognized, its intensity can be reduced more rapidly than normal.

Environment classification

Most advanced digital hearing aids can classify the current listening environment into one of the several pre-defined acoustic environments. These pre-defined environments usually include speech in quiet, speech in noise, noise alone and music. Noise can also be further classified into noise types, most notably wind noise versus others. Classification is based on numerous parameters which include overall level, spectral shape, depth and rate of modulation and co-modulation channels.

The result of the classifier is used to automatically enable/disable other features in the hearing aid which includes the directional microphones, adaptive noise reduction, and wind noise reduction. Enabling / disabling any feature usually changes sound quality and it could be disconcerting to the user if the sound quality is to change if the world around them has not changed substantially. Environmental classifiers take a cautious approach, analyzing the environment over may seconds or several tens of seconds, before pronouncing their conclusion.

Automatic telephone detection

One important listening situation of importance is talking on the telephone. Hearing aids can detect if a strong external static magnetic field permeates the hearing aid. This could result in the hearing aid automatically switching to telecoil every time the wearer holds a phone to the ear. Telephones do not always produce a sufficiently strong static magnetic field. Things other than telephones can produce strong fields.

Automatic detection can be made more reliable by a small circular magnet stuck onto the telephone. The ability of hearing aids to reliably recognize the close presence of a telephone is helped by a wireless link between the hearing aids. Presence of a telephone can also be detected by a change in the feedback path.

When room loops are used they don't produce a static magnetic field and do not cause the acoustic feedback path to change, so to listen to a room loop through hearing aids in which the telecoil can only be selected automatically, a magnet should be suspended beside the hearing aid.

Data logging features in hearing aid

Hearing aids are capable of storing information about the listening situations they are used in. They can also store information pertaining to how often it was used in each situation, how the client adjusted the hearing aid in each situation, and how the hearing aid self adjusted or activated signal processing features in each situation. This can be accessed by the clinician during subsequent visits. It helps the clinician to adjust the hearing aid in order to make the amplification acceptable to the user.

Uses of data logging

1. If the logged data shows the user is constantly turning the volume control up / down in particular listening situations, the clinician can make this adjustment permanently.

2. If the user has difficulty manipulating the hearing aid, the logged data will indicate whether the client is actually using the hearing aid.

3. If the user complains that the batteries don't last long enough, the daily use patterns will show whether the user is turning the hearing aid off after removal, or whether there are any problems with the batteries.

The software available with fitting systems can automatically analyze the downloaded data and recommend specific changes to the hearing aid settings.

Trainable hearing aids

Commonly hearing aids are fine tuned by the clinician based on feedback received from the user. It should be pointed out that the user could have limited experience listening to a restricted range of sounds. Commonly the user feedback depends on the listening experiences in the first few weeks of hearing aid usage. The impressions of the user could very well change as experience is gained with amplified sound. These problems can be better handled if the hearing aid is trainable and they are also known as self-learning hearing aid. In the trainable hearing aid, when the user adjusts a control / controls which not only cause the amplification to change, it remembers the setting for future use. In a simple trainable hearing aid, the device just remembers and averages over time the position of the control that the user prefers. When the hearing aid is turned on next, the amplification characteristics are already those corresponding to the average amplification characteristics (i.e gain) previously preferred by the client.

Sophisticated trainable hearing aids remember not only the position of the control preferred by the client, but also some aspect of the acoustic environment in which the control was adjusted. After accumulation of training history, the hearing aid can infer what position of the control are preferred by the client in different situations.

Some advanced hearing aids are capable of carrying out self-learning calculations and subsequent calculations separately in each of a small number of environmental categories (speech in quiet, speech in noise, noise only, music). The sound world is not intrinsically categorized. A better approach would be for a hearing aid to use the training preferences to deduce the relationships between one / more amplification characteristics and one or more aspects of the environment.

One example would be the volume control of the hearing aid. The data available to the hearing aid after the user has made adjustments to suit the listening needs of varying environments. Using these data the hearing aid can automatically adjust the volume requirements as per the environmental needs. Users can also adjust 2 or 3 controls in order to achieve a preferred tonal quality and loudness.

Active occlusion reduction

The user's own voice could create excessive low frequency sound in an occluded ear canal. A signal processing solution to this problem is to sense the occlusion induced sound in the ear canal, invert the sound pressure, and output this inverted sound

wave through the receiver back into the ear canal. Since the original and re-introduced sound have opposite polarity, they cancel out each other resulting in a sound pressure level much lower than the generated sound. This is similar to the principle used in active noise reduction headsets. In order to achieve this effect additional microphone sensing the ear canal sound pressure, the internal negative feedback loop and various filters are needed.

Image 1.71 shows the essential ingredients. Filters A and B are needed to ensure that the loop containing them and the two inward looking transducers provide negative feedback throughout the frequency range for which there is appreciable gain around the loop. The negative feedback loop reduces the level of any signal entering the loop.

Active occlusion reduction has a number of advantages over passive occlusion reduction. They include:

1. The earmold can be closed or have a small vent, so leakage out is greatly reduced, which makes it possible to achieve greater high frequency gain without feedback oscillation occurring.

2. The active system reduces any sound entering the ear canal, including sound that travels in through the vent. Consequently, amplified sound can become dominant over vent-transmitted sound down to a very low frequency. This means that directional microphones and adaptive noise reduction systems can work over the entire audio frequency range, rather than just the more restricted frequency range over which amplified sound usually dominates vent-transmitted sound. This is an advantage because noise is often more dominant at low frequencies than at high frequencies.

Disadvantages of active occlusion reduction

1. The extra microphone needs additional space in the ear canal making the hearing aid slightly bigger. This device hence cannot be used in individuals with small ear canals.

2. The port for the extra microphone opens into the ear canal, creating further place for wax and moisture to enter the hearing aid causing it to fail. An effective wax barrier is hence essential.

3. The additional signal processing, which must operate at a high rate to avoid delays around the feedback loop, uses additional battery current thus shortening the life of the battery.

Own voice detection

Many signal processing schemes rely on estimation of the SNR to approximately adjust the amplification characteristics. The close proximity of the hearing aid to the mouth and its location somewhat above and behind the mouth causes the input to have a particularly high level, and a pronounced low-frequency dominance whenever the aid wearer talks. This can mislead the algorithms if their job is to optimally adjust the amplification to allow the wearer to hear and understand other people. Hearing aids can use the high level low-frequency dominance, and equal input signals at both hearing aids (since mouth is located midway between the ears) to infer whether the dominant speech signal is originating from the aid wearer or from another person. If the hearing aid is provided with active occlusion reduction, the additional internal microphone provides a further information source that makes the task of own-voice detection easier and accurate.

Self checking hearing aids

The additional microphone sensing sound within the ear canal can also be used to check for proper performance of the hearing aid. A fault in the electronics, a receiver blocked with wax, or a very poorly fitting earmold will all cause the sound in the ear canal to differ from what should be there given the input sound and the amplification characteristics programmed into the hearing aid. An automatic comparison of the expected and actual sound characteristics can be used to trigger an audible warning to the user that the hearing aid is faulty and needs servicing.

Predictive bandwidth extension

Speech sounds with significant energy present in the high frequency range (4-8 kHz) also tend to have significant energy in very high frequency sounds, above 8 kHz. Hearing aids can use this correlation to artificially extend the bandwidth to 12 kHz provided the receiver has the capability to output audible sound over this extended range. Improved speech intelligibility, naturalness, or localization would also depend on the user having sufficient remaining hearing ability to detect and analyze sounds in the extended frequency range.

Self fitting hearing aids

If a automatic audiometer is built into a hearing aid, the hearing aid itself can measure the hearing thresholds of the user, apply a prescription formula to calculate the appropriate real-ear response, and adjust the hearing aid to approximately match this response. This process is supposed to mimic a skilled clinician. Studies reveal that both the automated threshold measurement and the automated hearing aid adjustment can potentially be as accurate as the same process carried out by a clinician.

One obvious concern could be that patients may do the threshold testing in a noisy environment as this invalidate the threshold values. The hearing aid can use the normal microphone to monitor noise levels, and even the noise spectrum during the testing and can advice the patient to move to a quieter place if the noise levels measured are too close to the thresholds obtained.

Such a device should not require connection to a computer. The patient themselves can carry out fine tuning if the hearing aid also incorporates a trainable algorithm.

One of the major difficulties in evaluating novel processing schemes is allowing for the effects of familiarity and practice. If sound is markedly altered by the processing, it is likely that experimental subjects will need considerable listening experience, and perhaps even systematic training before they can use the altered / new cues to identify spoken words. Extensive listening experience is difficult to provide inside the lab. One method to minimize this dilemma is to test for speech discrimination which is the ability to differentiate contrasting sounds.

All the processing algorithms will have to be adjusted to best suit each patient. It should also be stressed that by far the best way to improve intelligibility of speech is to remove all noise and reverberation from the signal before presenting it to the hearing impaired person. One solution to effectively reduce noise and reverberation is to use directional microphones.

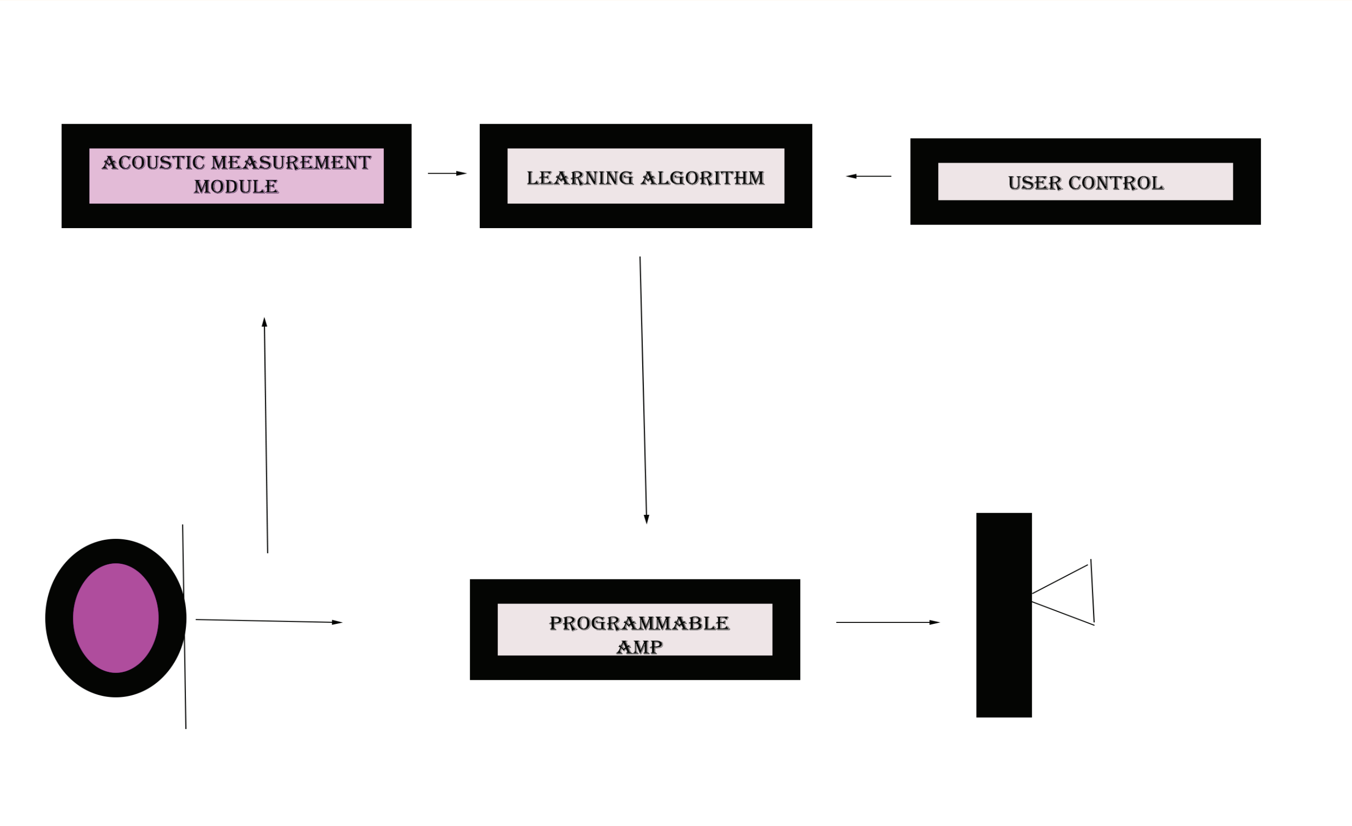


Image 1.70 showing a block diagram of trainable hearing aid

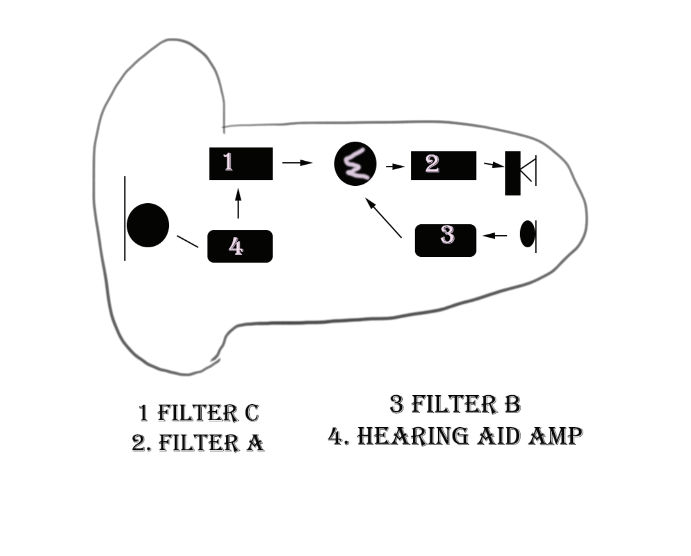


Image 1.71 showing active occlusion reduction system inside an ITC hearing aid.

Hearing Aid Candidacy

The decision to try hearing aids ultimately rests with the user. Many of hearing impaired patients would look up to the clinician to make a recommendation. Before recommending the clinician should take into account many factors and not the pure tone audiometric threshold of hearing alone. The initial motivation to secure hearing aids has been demonstrated to be the key determinant of whether patients would continue to use the hearing aid. The motivation reflects the balance of all the advantages a patient expects hearing aids to provide offset by all the expected disadvantages.

The advantages expected by the patient are affected by the degree of disability they feel they have. The advantages and disadvantages expected by the patient are affected by what the patient has been told about hearing aids by others. Disadvantages potentially could include the impact of the hearing aid on the self image of the patient. Even though hearing aids help the patients to hear both in quit and noisy environments, they help more in quite environment.

A clinician who encounters a hearing impaired person who does not want hearing aids, the clinician should first find out whether this is because the patient is not aware of the loss / or the difficulty, or does not wish to wear hearing aids. The exact reasons for the refusal should be discovered.

Difficulty in managing the hearing aid can impact its use. Patients with tinnitus when using hearing aids would find that it diminishes the problem. Tinnitus can positively affect the candidacy for the hearing aid. Presence of central processing disorders and extreme old age can affect the candidacy. Persons with a severe / profound hearing loss are likely to receive more benefit from cochlear implants than from hearing aids

To fit / not to fit is a difficult question that needs to be answered by both the patient and the treating clinician. The final decision to try out the hearing aid is the decision of the patient, but of course the clinician can greatly influence this decision.

Incidence of hearing loss is very common. Approximately 10-16% of adult population will have a four frequency average hearing loss in the better ear of greater than 25 dB. Majorty of patients with hearing loss don't need hearing aid. When hearing loss is based on self reported difficulty, only the older hearing impaired persons are likely to seek help.

Factors influencing the use of hearing aids

Many attempts have been made to calculate the degree of pure tone hearing loss that would distinguish those who will benefit from the use of hearing aids from those who will not. These attempts have largely been unsuccessful.

1. Attitude & motivation

The strongest of the factors that affect the use of hearing aid happens to be the attitude of the user and their motivation level. These two factors are the result of accumulation of a variety of positive and negative factors like:

Acknowledgment of loss: Important question that a patient needs to answer is whether he / she realizes that hearing is not normal. The patient should fully acknowledge the presence of hearing impairment.

Communication needs: This is another important factor that could push a person to seek medical help. Individual with hearing loss would need

to concentrate more to understand the spoken word thereby causing a lot of fatigue which affects their performance at work.

Consequences: Does the hearing loss cause the patient to refrain from activities that he / she would otherwise like to do, or does it cause the patient to have negative feelings about life? How much of participation restriction (hearing handicap) does the patient have? How restricted the patient feels? Honest answers to these questions will enable the user & clinician to arrive at a honest decision whether to go in for a hearing aid or not.

Self image: The thought of loss of self esteem that wearing a hearing aid could cause could be a deterrent for using hearing aids.

Expected benefit: This is another factor that could tilt a person's decision either way.

Fear / uncertainty: Anticipation by the patient of the difficulty in understanding how to operate hearing aids could be a factor in the decision making.

Cost: Cost of the device is another factor that could play a role in the decision making process.

Influence of others: Majority of users of hearing aids are positively influenced by their family members to obtain hearing aids.

Panels have been designed to decide which patients would benefit from the use of hearing aids. Panel outlines six tools that could provide insights into a patient's attitude towards wearing hearing aids. Many clinicians prefer to talk to the patient, but unless the conversation covers the main topics addressed in these tools it may not be useful at all.

Saunders et al 1997 recommends asking two key questions in order to quickly determine the motivation level of the patient.

1. What prompted you to come for a hearing test?

2. What do you expect to gain from this visit?

WANT: (The wishes and needs tool)

This tool includes two simple questions that reflect motivation and which have been shown to relate to the success of the fitting.

1. Overall, how much difficulty do you have in hearing?

2. How interested are you in obtaining hearing aids? This question is best asked after the patient has been provided with information about his or her loss and the likely benefits of hearing aids.

HASP (Hearing aid selection profile): Questionnaire has about 40 questions which the prospective user needs to answer. It would provide the clinician some useful insights on the willingness of the patient to go for a hearing aid. The questionnaire can be downloaded from here.

ALHQ: The attitudes towards loss of hearing questionnaire comprises of 22 questions which provides a score for each of them.

Denial of hearing loss - does the patient consider

he / she has a problem with hearing?

Negative associations - stigma related concerns.

Negative coping strategies - fear and avoidance of communication.

Manual dexterity and vision - Impacting on ease of handling small objects.

Hearing related esteem - Has hearing loss impacted on confidence of the individual.

SPHA: The self perception of hearing disability asks patient on a scale from 1-10, 1 being the worst and 10 being the best, how would you rate your overall hearing disability? Those who answered 6 or less were more likely than not to acquire hearing aids. Almost everyone who gives a rating of 1 or 2 acquires hearing aids and almost no one who gives a rating of 9 or 10 does.

HARQ: The hearing attitudes in rehabilitation questionnaire comprises of 40 questions assessing attitudes towards impairment, loss associated stigma, loss minimization, aid associated stigma, acquiring hearing aids, pressure from others and expectations.

2. Hearing impairment

The hearing loss impairment of the patient influences, but by no means determines how much impairment, activity limitation, and participation restriction the patient believes he/she has. Degradation of frequency selectivity, temporal resolution and spatial processing ability are only partially correlated with the loss of sensitivity revealed by the audiogram. Despite these pitfalls PTA is a good indicator of the overall degree of physiological impairment, is a moderate indicator of activity limitation, but is a poor indicator for participation restriction.

It should also be stressed that the fact the patient has come to the clinic seeking help at the behest of others is no bar to successful hearing aid use. Irrespective of the combination of factors that influences patients, it is their own acknowledgment of hearing disability and the motivation to do something about the problem are strong predictors of how much they use hearing aids.

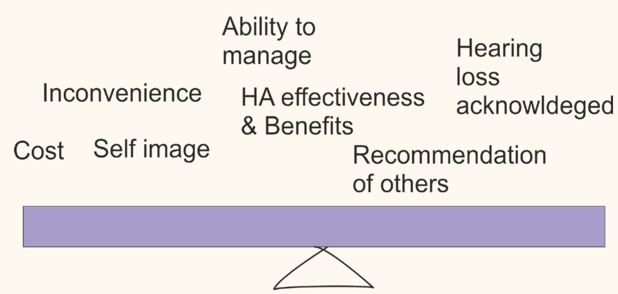


Image 1.72 showing visual representation of health brief model. Factors listed on the right could encourage a person to acquire hearing aids while factors on the left side are more likely to discourage them.

2 Pure tone loss and audiogram configuration

The benefit that people obtain with hearing aids, the proportion of hearing impaired people who use hearing aids, and the number of hours per day that people use hearing aids all increase with the degree of pure tone hearing loss. For mild to moderate hearing losses, hearing aid usage is not

common. The effect of hearing loss configuration on usage of hearing aids is unclear. Disability is closely related to low frequency hearing thresholds than to high frequency thresholds, although both frequency ranges are important for good hearing.

It has been suggested that people with three frequency average losses (500, 1000, 2000 Hz) of less than 25, 30 or 35 dB will not benefit from hearing aids, but persons with greater losses will. There is also no data to suggest that pure tone loss is a reliable indicator of who will benefit with the use of hearing aids.

In the event that pure tone thresholds are used as a guide to candidacy for hearing aid, hearing thresholds in the ear with the larger pure tone loss should be used,as they appear to better predict hearing aid candidacy than loss in the better ear in persons with mild / moderate hearing loss in both ears. For the sake of convenience, a two part criterion for hearing aid candidature can be used which include, four frequency average loss greater than 35 dB in better ear or greater than 45 dB in the worse ear when combined with a difference of 15-35 dB between the ears.

Special issues faced by persons with ski-slope hearing loss

1. Persons with good low-frequency hearing inappropriately consider that they do not have hearing problems. This is more likely if the hearing loss is gradual. In these patients motivation will play a huge role in hearing aid acceptance.

2. The potential benefit from a hearing aid is least for extremely steep losses and where the high frequency loss is the greatest. The wider the frequency range where the loss is between 20 and 80 dB HL, the greater will be the benefit of the hearing aid.

3. Occlusion effect will be a problem that must be dealt with at fitting, almost certainly with an open fitting. Now a days closed fitting occlusion canceling devices are available.

4. A hearing aid that provides gain for only high frequencies may improve clarity, but have very little effect on the loudness.

Although Pure tone hearing loss is an obvious predictor of candidacy for hearing aid use it is unreliable when used as a sole indicator except in cases of normal hearing (no benefit) and severe hearing loss (substantial benefit). For all the people with losses in between, it is best to use hearing thresholds as a guide for further questioning.

Patients with moderate hearing loss will not be able to hear parts of speech signal in most listening situations. If they state that they experience no disability and hence do not wish to obtain hearing aids, the reason should definitely be investigated.

3. Speech identification ability

People with poor speech identification ability in quiet are more likely than people with higher speech identification ability to use hearing aids. High speech scores do not prove that a person's hearing is too good to benefit from hearing aids.

Conversely patients with the poorest speech reception thresholds in noise report the least benefit from amplification. Predicting candidacy of an individual from speech identification ability is really not a valid solution, because speech scores depend strongly on test conditions, such as speech level, noise level, reverberation, spatial arrangement and distance off the sources, and difficulty of

the speech material. Any conclusion of a person having no hearing problems would be applicable only to the conditions under which the speech measurement was performed.

4. Self reported disability

Persons who seek hearing aids are likely to be aware of the hearing disability than people who do not. This is because it causes limitations of activity, and participation restriction in these individuals. Patients who initially report the most disability are the most likely to report that they are helped by hearing aids. Persons who completely accept that they have hearing loss use hearing aids more than those who grudgingly accepted their need for amplification.

A severe visual disability reduces the ability of the person to lip read. In noisy places lip reading is very important and helps the individual to compensate for the hearing loss. Loss of vision would make the loss of hearing to become more worse because of this vital factor. Vision loss could also make it difficult for the person to manage hearing aids as the controls are very small.

5. Acceptance of noise

Hearing aids amplify every sound. Excessive amplification of background noise is the common complaint of hearing aid users. Studies reveal that environmental sounds bother people more when hearing aids are used than when listening unaided. Persons who are willing to listen to speech at very poor SNRs appear to be more likely to use their hearing aids more than those who require a more positive SNR.

A test has been developed to assess the SNR that people need for speech to be acceptable to them. This test is known as the ANL test (Acceptable Noise Level). It is performed by first having the patient adjust speech to the Most Comfortable Level (MCL). Noise, like speech babble is then added and the patient is asked to adjust it to the highest level that the patient can accept / put up with while following the story narrated in the original speech. The level selected is known as the Background Noise level (BNL).

ANL is calculated as MCL minus BNL. It is the poorest SNR that is acceptable to the patient. Persons with small ANL (>7 dB) are likely to become full time users of hearing aids because they are able to put up with amplified noise close to the level to the signal of interest. Persons with large ANL (>13 dB) are likely to use hearing aids less or not at all because they find amplified noise objectionable in too many situations.

ANLs can be obtained in the sound field or under headphones, with all frequencies amplified by the same amount, or with frequency dependent amplification prescribed on the basis of the audiogram. ANL values are affected by the competing signal used.

ANLs are not related to gender, hearing thresholds, loudness discomfort levels, acoustic reflex thresholds, or contra lateral suppression of otoacoustic emissions and are only weakly correlated to age and speech intelligibility scores in noise. Lack of gender difference in ANL occurs despite males having higher MCLs and BNLs than females.

6. Listening environment, needs and expectations

Hearing aids provide much more benefit in some situations than in others as in the case of listening to soft spoken person in a quiet place, and loudly spoken person in a noisy, reverberant place. Peo-

ple who are new to hearing aids would expect that they will be beneficial in noise as they are in quite. They also have a tendency to underestimate the negative aspects of hearing aids, such as feedback oscillation and amplification of background noise. It is rather important to identify those patients with unrealistic expectations in either direction before deciding on hearing aid trial in order to alter their expectations to more realistic levels. This can be achieved by counseling or by demonstration of the sound quality that is possible in different situations using recordings and computer simulations.

Any person with a hearing loss will have more difficulty hearing in both very quiet and very noisy situations than a person with normal hearing. If a patient initially reports difficulty in any one of these situations, then it is worthwhile questioning in order to find out if the other type of situation ever causes problems. The more a person is in contact with other people the more likely it is that the advantages of hearing aids will outweigh their disadvantages. A loner with a loud TV and radio does not need a hearing aid.

Another provision is the use of directional microphone or dual microphones that can be selected to function as a directional microphone, significant benefit can be experienced even in noisiest places. When noise and the wanted signal are coming from different directions, hearing aids with directional microphones may allow the patient to communicate in poorer signal to noise ratios than would be possible without hearing aids. The directional microphone will decrease the amount of noise being perceived, and increase the SNR, by an amount that depends on the specific listening situation. Persons with mild hearing losses usually would appreciate the physical comfort and freedom from occlusion than an open fitting gives. Hearing aid is hence not directional in low frequencies. The directional microphones will help for only those frequencies high enough to be unaffected by the open ear canal, and low enough that it is noise limiting audibility rather than hearing thresholds.

The quantum of benefit that omni-directional and directional hearing aids provide to people with mild/moderate hearing loss can be calculated using the speech intelligibility index calculation method. The hearing impaired person has the maximum difficulty, both absolutely and relative to normal hearers in noisiest places. The hearing aid whether in omni-directional or directional mode, helps least in noisiest places. As input level increases, the gain of the hearing aid decreases which decreases the frequency range over which the directional microphone can improve SNR. The hearing aid, most needed in noise, most helps in quiet environment.

7 Stigma and cosmetic concerns

Patients are often concerned about the appearance / visibility of hearing aids because it could prove to be a social stigma. Many adults are concerned that they will be perceived as being older if they wear hearing aids. This concern is understandable as hearing aids are commonly used by people in older age group. Older persons are more willing than youger persons to adopt hearing aids. Studies reveal that stigma linking hearing aids with age is the major reason for the limited penetration rate of hearing aids.

External stigma - Stigmatization by others. This is actually not a strong phenomenon. Patients begin to realize this only after they start using hearing aids. It should also be pointed out that hearing aid users actually have a more positive self image

than hearing impaired persons who don't wear hearing aids. The hearing aid candidates should be reassured that they will not be viewed negatively if they wear hearing aids.

Size of the hearing aid - This of course matters. Users would pay more for a less obtrusive hearing aid. People choose smaller hearing aids over larger ones if choice is given to them. Many users feel that small and suitably colored hearing aid can overcome cosmetic issues to some extent.

8. Manipulation & Management

Operating hearing aids can be daunting for old individuals who have tremors. Finer movements of fingers are found wanting in this group. Manipulation difficulties can be caused by poor flexibility of finger / arm joints or by low tactile sensitivity. Low cognitive functioning can also prevent patients from properly using their hearing aids. The miniature hearing aid could make insertion of battery and operating the volume control and on and off switch rather difficult. Insertion of the hearing aid into the ear canal also could prove difficult and a task that needs to be learnt. Management difficulty increases with each decade after the patient has passed the age of 60. Patients who have tasted the usefulness of hearing aid could continue to use them despite manipulation difficulties. Ease of management is a critical issue in hearing aid selection and patient instruction. The older a patient is, the more likely it is that they will regard ease of management as the most important factor in hearing aid selection.

In patients with issues pertaining to manipulation and management issues assisted listening devices can be preferred. They could be infrared / radio frequency devices for TV listening, body worn hearing aids, or hand held devices with output via headphones could be considered.

10. Age

Age (either old age / infancy) by itself does not directly affect candidacy for hearing aids. It can of course affect other factors like manipulation difficulties, cosmetic preferences, hearing needs, hearing impairment. This can have an indirect effect on candidacy. Increasing age also increases the likelihood of an auditory processing disorder and a need for greater SNR in noise.

Illness accompanies old age. Persons with hearing loss could regard their health problems as more pressing than that of their hearing problems. The prevalence of dementia also increases with age.

11. Personality

Several personality trains are associated with a greater likelihood of take-up for rehabilitation and a greater degree of self reported benefit from rehabilitation with hearing aids.

Internal locus of control - Patients who feel that they control the things that happen to them are more likely to acquire and use hearing aids than those who feel that things just happen to them (an external locus of control). Closely related to an external locus of control is learned helplessness. Those persons who strongly believe that their lives are controlled by others are also more adversely affected by loud sounds.

Extroversion - Patients with outward looking personality report more benefit from amplification and less activity limitation and participation restriction when aided than patients with inward looking personality.

Agreeableness - Patients who are rusting, sympathetic and helpful benefit from hearing aids.

12. Central auditory processing disorders

As age increases, so is the likelihood of hearing loss due to decrease in the central auditory function. These persons will be susceptible to interference from competing signals. Studies have revealed that an auditory processing disorder diminishes the use, benefit, performance that hearing aids provide. These patients can be identified by dichotic speech test. One form of central auditory processing disorder is known to be present commonly in persons with sensorineural disorders (spatial processing disorder). This appears to be a contributor to the deficit in SNR experienced by hearing impaired persons.

The presence of an auditory processing disorder should hence not prevent the clinician from fitting hearing aids. The presence of a central processing disorder may however explain why some people report little benefit from hearing aids. Since central processing deficits can appear / increase in magnitude as persons age, it is possible for hearing aids to become less effective with time. When this occurs these persons would complain of the hearing aid output becoming distorted even though the hearing aid electro-acoustic performance remain unchanged.

Wireless system provide a solution to the problems caused by central auditory processing disorders because of their ability to greatly attenuate unwanted signals and noise. A proportion of these patients would use these systems regularly despite the logistical difficulties associated with their use.

13. Tinnitus

Many persons with hearing loss also have tinnitus. Amplification of external sounds could often relieve the adverse effects of tinnitus including its psychological effects. Amplified sounds can also provide partial or even complete masking of the tinnitus. Since tinnitus is often best masked by high frequency sounds, open fittings can be an effective way to provide the amplification needed for both speech intelligibility and tinnitus masking. A combination of tinnitus retraining therapy, open fit hearing aids during waking hours, and a noise maker (sound enrichment) device near the bed during sleep has shown to markedly reduce tinnitus problems for patients with mild sloping hearing loss.

Commonly hearing aids include an optional controllable internal noise source so that tinnitus masking is not dependent on serendipitous internal noise or on amplification of noise in the environment. The presence of tinnitus increases the likelihood that a person will accept hearing aids, because of the masking that amplification provides and should hence be considered a positive factor when assessing hearing aid candidacy.

14. Factors in combination

A difficult audiogram (ski-slope loss) does not rule out the use of hearing aids. Neither does difficulty manipulating hearing aids, a belief that help is needed only in noisy places, poor speech discrimination, concern about the appearance of the hearing aid, arthritic fingers nor a slightly hesitant attitude to trying out hearing aids. The patient in whom all of these are true is less likely to find hearing aids useful than would a patient for

whom there was only one of these difficulties to overcome.

The clinician's duty is to identify, for each patient all the potential obstacles to success, overcome those that can be overcome and weigh up the remaining difficulties against the likely benefits that the patient will receive.

Upper limit of hearing loss that can be covered with hearing aids

Advent of cochlear implants have really changed the scenario in such a way that even patients with profound hearing loss (which could be below the level that can be amplified by a hearing aid) can also be rehabilitated. Clinicians while managing paients with severe / profound hearing loss can use the following combination of management modalities:

1. Unilateral / bilateral cochlear implants

2. Bimodal fitting comprising of a cochlear implant in one ear and a hearing aid in the other.

3. Hybrid fitting, comprising of a cochlear implant and hearing aid in one ear, combined with an implant, a hearing aid, another hybrid fitting or nothing in the other ear.

4. Unilateral or bilateral hearing aids

5. Tactile hearing aid

Poor speech identification ability

Word recognition scores, obtained using headphones, of less than 50% indicate that hearing aid benefit will be limited to help with lip reading, monitoring one's own voice and detecting environmental sounds. Speech identification scores obtained using headphones are not a good indicator of whether a person will benefit from hearing aids because a hearing aid does more than just amplify. It reshapes the speech spectrum relative to the flat frequency response that is available within the audiometer. Speech scores obtained with an individually prescribed hearing aid are usually greater than those obtained without frequency shaping, and often they are much greater.

Speech identification scores have an inherent random component. As the number of test items used decreases the spread of scores from test to retest widens. Speech scires also depend on the level at which they are presented. The test hence should be performed at several sound levels to improve its reliability. A lot of these problems can be rectified by spending a lot of time testing, and by performing the testing with an amplification system that has a gain-frequency response appropriate to the patient. The main fundamental problem would be in determining what cut off score separates hearing aid candidates from non-candidates. Some persons with profound hearing loss wear hearing aids as they help with lip-reading and also because they give the user an awareness of sounds in their environment. This decreases their stress, tension and insecurity. The increase in intelligibility offered by hearing aids is highly dependent on the levels of speech and noise used.

What to use Hearing aids / cochlear implants?

Factors affecting the performance level of Cochlear implants:

There are many factors that should be considered

before a person can receive a cochlear implant. The basic requirement would be a reasonable expectation that the cochlear implant would provide speech identification ability superior to that which can be achieved with hearing aids. Since there is a wide range of speech identification ability provided by cochlear implants compared to that of hearing aids it would serve as a boon to patients even with profound hearing loss. Better implant performance is likely in patients with:

1. Severe / profound hearing loss for the shortest time

2. Patients who are implanted are as young as possible and in the case of children preferably before the first birthday.

3. Who had hearing at the time they were acquiring language

4. Had the least hearing loss, and the highest speech intelligibility scores with hearing aids before implantation

5. Who can be motivated to engage in rehabilitation activities

Predicting cochlear implant candidacy from hearing aid thresholds

There is a wide range of speech identification performance across people with the same degree of hearing loss who wear hearing aids. Prediction of cochlear implant candidacy from hearing aid thresholds could be very dicey. One can of course ask how much speech intelligibility is conferred on an average by cochlear implants compared to that of hearing aids for various degrees of hearing loss. Adults with cochlear implants typically have better speech identification ability than is typical of adults whose three or four frequency average hearing thresholds average 80-85 dB HL when wearing hearing aids. This means on an average cochlear implantee outperforms the average hearing aid wearer with this degree of hearing loss.

Predicting improvement in speech scores following cochlear implant when compared to hearing aid speech scores

Better prediction of implant benefit can be made if actual speech performance with hearing aids is known already. For adult patients with a long standing hearing loss there will be ample opportunity for the patient to have been fitted with hearing aids, for the hearing aids to be fine tuned to get the best possible performance and for speech identification ability to be measured.

Commonly applied criterion for implantation is that for adults, open-set speech sentence scores in quiet with hearing aids should be less than 50% in the ear to be implanted. This simple criteria does not take into account the very important factor of how long the person has had profound hearing loss.

A less strict criterion that can be applied is for a patient with a progressive hearing loss in whom deafness is expected to worsen. It is not uncommon for pre-lingual deaf adults to achieve open-set speech scores that are no higher than those obtained prior to implantation.

Many successful implant users report that:

1. They use the device regularly.

2. They are satisfied with it.

3. It helps them to monitor their own voice

4. It facilitates their independence and employment

5. It also enables them to detect and recognize environmental sounds which increases their feeling of security.

Cochlear implants are more effective in conveying mid and high frequency sounds than low frequency sounds. They are also more effective at conveying information about spectral shape than conveying information about fine time pattern and pitch of sounds. Reasons for the failure to accurately represent pitch are not fully understood, but could be related to the inability of individual electrodes to stimulate just low-frequency neurons. On the other hand hearing aids are usually more able to convey pitch and other low frequent information than to convey higher frequency spectral information because of the fact that residual hearing is usually present in the low frequencies. Hearing aids and cochlear implants provide information complementary to each other.

Bimodal stimulation

When hearing aid and implant are used in opposite ears, the combination is referred to as a bimodal fitting. In nearly all cases the implant alone provides considerably better speech intelligibility than the hearing aid alone. The combination of these two always provide better speech intelligibility in noise rather than implant alone. The added benefit of hearing aid increases for many months following implantation.

Major role of the hearing aid is in providing low frequency information. It is possible that excessive mid and high frequency output from the hearing aid can decrease speech intelligibility by providing cues that conflict with the implant. Benefit happens to be the best when the speaker is positioned nearer to hearing aid ear.

Hybrid devices

When a hearing aid and an implant are in the same ear, the combination is referred to as a hybrid that provides the wearer with electroacoustic stimulation. In a hybrid device signal components in the incoming sound above a certain cut-off frequency are presented via the implant and signal components below the cut-off frequencies are presented via the hearing aid. These devices are most suitable for persons with steeply sloping audiograms. Mild to moderate loss in low frequencies enables the ear to make good use of low frequency sounds delivered by the hearing aid, including the perception of pitch, whereas the severe or profound hearing loss in the high frequencies is best served by the implant. One method used to select the cut-off frequency is to select the frequency at which the audiogram equals about 70dB HL. Since the implant does not have to convey low frequency signal components, there is no need to deeply insert the implant electrode to reach the low frequency signal part of the cochlea.

Hybrid devices provides better speech intelligibility in noise than either device by itself could. This features ensures better perception of music.

Tactile aids

Some deaf persons receive only vibratory informa-

tion from hearing aids. This provides just the time and intensity information with little or no spectral information. These persons could benefit from purpose designed vibrotactile / electrotactile aids. These aids can encode more speech information into the sense of touch than is accidentally encoded by hearing aids designed to provide an acoustic stimulus. All commercially available tactile aids use vibration as the stimulus. These aids are capable of creating small discharges in response to sound which could be passed on to the skin. This would be perceived as tactile sensation.

These tactile aids provides information that could supplement lip reading, but they provide much less information than a multichannel cochlear implant.

Medical contraindications for hearing aid fitting

Any condition that would cause a clinician to refer the patient for medical assessment will temporarily / permanently halt the process of hearing aid fitting. They include:

1. Hearing loss of sudden onset

2. Rapidly progressing hearing loss

3. Pain in either ear

4. Tinnitus off sudden onset, or unilateral tinnitus

5. Unilateral or markedly asymmetrical hearing loss of unknown origin

6. Vertigo

7. Headaches

8. Conductive hearing loss of any origin

9. Otitis externa / otitis media

10. Cerumen filling more than 25% of the cross section of the ear canal

11. Atresia of external auditory canal

Hearing Aid Prescription

Prescriptions for hearing aid is usually given following a formula that links some characteristics of a person to the target amplification characteristics. Prescription formulae most commonly used are based on hearing thresholds, but sometimes supra-threshold loudness judgments are also used.

Some of the commonly used prescription procedures for linear hearing aids include:

POGO

NAL

DSL

For all of these, gain can be prescribed based on hearing thresholds alone. These formula all contain variations of the half-gain rule, these variations could be very different that the resultant prescriptions could differ greatly, that too in persons with a sloping hearing loss. For non-linear aids all available prescription procedures should include some aspect of normalizing the loudness of supra-threshold sounds.

Procedures like LGOB, IHAFF, CAM-REST and FIG6 aim to normalize loudness at all frequencies, at least for sounds with levels above the compression threshold of the hearing aid.

Concepts behind a prescriptive approach of hearing aids

Hearing losses vary widely in their severity, configuration and type. Hearing aid should hence be selected, its amplification characteristics adjusted to be appropriate for each hearing impaired person. Only way this is possible is to follow a prescription procedure. This procedure involves measurement of the hearing defect in an individual, and the amplification characteristics that are required are usually calculated from this value. The amplification characteristics required are referred to as the amplification target / prescription target. The measured characteristics nearly always include hearing thresholds, and very often these are the only characteristics that are measured.

Evaluative approach

In contrast to the prescriptive approach, this approach is purely hypothetical. In this approach, a number of hearing aids or response shapes would be chosen randomly and then tested on the hearing impaired person to find the best suited one. This approach could be totally impractical in its purest form because of the sheer number of potential amplification characteristics that needs to be evaluated.

Prescriptive procedure for selecting a hearing aid has a long history. In 1935, Knudsen and Jones proposed that the gain needed for each frequency was equal to the threshold loss at the same frequency minus a constant. This is also referred to as the mirroring of the audiogram, because the shape of the gain-frequency curve equals the inverse shape of the hearing loss. With mirroring procedures, every 1 dB increase in hearing loss requires 1 dB of additional gain to compensate. In sensorineural hearing loss, the gain needed to restore normal loudness perception is equal to the threshold loss only when the person is listening at threshold. For all higher levels, this amount of gain would be excessive. Mirroring hence leads to excessive gain, especially for those frequencies with the greatest hearing loss.

Another development is to base the gain needed on the person's Most comfortable level rather than

on their thresholds. Watson and Knudsen suggested that speech should be amplified sufficiently to make speech energy audible and comfortable. Their specific formula involved Most comfortable Level (MCL), but surprisingly did not take into account the variation of speech energy across frequency. Lybarger in 1944 made a very important observation i.e. averaged across frequency, the amount of gain chosen by people was approximately half the amount of the threshold loss. This is known as the half-gain rule. This rule underlies several other prescriptive procedures.

The two methods that are commonly followed include raising speech to the Most Comfortable Level and the half-gain rule. They are of course two sides of the same coin. For persons with mild and moderate sensorineural hearing loss the threshold of discomfort is a little different from normal. Most comfortable level is approximately half way between the threshold and discomfort so MCL increases by 0.5 dB for every 1 dB hearing loss. This explains why gain needs to be approximately half the hearing loss. If the aim is to raise the level of speech to MCL, then it is not possible to predict how much gain is needed at each frequency unless speech intensity at each frequency is taken into account. It should be pointed out that low-frequency components are more intense than the high-frequency components, the half-gain rule has to be modified. Either a little less low frequency gain has to be given or a little more high frequency gain or both.

The half-gain rule has to be modified further for severe and profound hearing losses. For hearing thresholds greater than 60 dB HL, discomfort thresholds are significantly above normal while Most comfortable level remains approximately midway between threshold and discomfort. This means that the MCL is elevated by more than half the hearing threshold loss. The gain hence must be more than half of the hearing loss. Two different auditory attributes could provide a basis for hearing aid prescription. One approach could be to measure some supra-threshold loudness percept like that of Most Comfortable Level (MCL), the second one being measurement of hearing threshold.

Prescription of gain has received far more attention than the prescription of maximum output despite the probable high important of maximum output for linear hearing aids. For non-linear (compression) hearing aids, now the most commonly used, the level at which hearing aids limit the maximum output is less important than for linear aids because some of the gain reduction that occurs when a linear limits is instead provided by the more gradual form of compression that commences at lower input levels.

Finding a simple relationship between hearing loss and gain has not been easy because:

1. The optimum gain frequency curve depends on the type of input signal, its level and spectral shape. Much of this research has been carried out only with linear hearing aids which may not be valid at present.

2. The optimum gain frequency curve may depend on things like supra-threshold loudness perception and frequency resolution ability in a way that cannot be predicted from threshold and may depend on other unknown factors.

3. The optimum gain frequency curve for a person may depend on the nature of the auditory input to which the person has become accustomed during the preceding months or years.

4. For a particular person listening to speech at a particular time and input level, there may not even be a single optimum gain frequency curve. The optimal value may depend on whether the person desires to maximize intelligibility or comfort or some other perceptual attribute of sound.

Prescription rules involve some formula. Once a prescription method has been chosen, the prescribed gains must be calculated. Formerly, this was done with tables, slide rules, or a calculator. Currently the formulae are included within the software produced by each hearing aid manufacturer for the purpose of adjusting their hearing aids. There are also standalone computer programs for hearing aid prescription procedures. Real-gain analyzers also include these formulae, so the measured real-ear gain easily can b e compared to the target gain-frequency curve it is meant to approximate.

Gain frequency response prescription for linear amplification

Linear hearing aids have the same gain-frequency curve for all input levels, until the output level is high enough to cause the aid to limit.

POGO (Prescription of Gain and Output)

This is a straight forward application of the half-gain rule with an additional low cut. The low cut was intended to decrease the upward spread of masking from low-frequency ambient noise. The low cut could be justified by the greater intensity of speech at low frequencies and by the lesser importance of speech information in very low-frequency region. The amount of low frequency cut specified would be based on the originator's experience. Insertion gain at each frequency is equal to half the hearing loss at that frequency plus constant from the accompanying table. This procedure is intended to be used only for hearing losses up to 80 dB HL.

NAL (National acoustic laboratories of Australia)

This prescription formula has been revised since its publication in 1976. The basic aim of the NAL procedure has been to maximize speech intelligibility at the listening level preferred by the aid wearer. Intelligibility was assumed to be maximized when all bands of speech are perceived to have the same loudness. If speech is too loud the patient will turn down the volume control. Decreasing the volume control setting will also decrease the loudness of all other frequency regions which may then become too low to contribute optimal to intelligibility. This calculation assumed that at all frequencies an extra dB of loss required an extra 0.46 dB of gain. This formula did not achieve equal loudness, especially in persons with steeply sloping losses. To solve this problem this formula was revised and became known as NAL-R. This formula retained the well established half gain rule for the three-frequency average gain.

DSL (Desired Sensation Level)

This formula aims to provide the hearing aid user with an audible and comfortable signal in each frequency region. It differs form NAL and POGO procedures in three different ways:

1. It prescribes a real ear aided gain rather than a real ear insertion gain.

2. DSL procedure is well integrated with measurement methods that are convenient for use with infants and young children without the use of average correction factors.

3. The DSL procedure does not attempt to make speech equally loud in each frequency region, although it does attempt to make it comfortably loud.

The DSL sensation level targets were derived and revised as given below:

1. In profound hearing losses, the desired sensation levels are based on the sensation levels experimentally found to be optimal.

2. For mild to severe hearing losses, the sensation level targets for bands of speech are placed one standard deviation below the estimated MCLs for pure tones.

3. For normal hearing, the desired sensation levels are those that are experienced by people with normal hearing when listening unaided.

This procedure uses desired sensation levels to calculate its target real-ear aided gain.

Difficulties encountered using prescription formula for hearing aids

Prescriptions for non linear hearing aids has several issues to contend with. Some of these issues include:

Acclimatization and adaptation to gain and frequency response:

Listening to amplified sound can produce gradual, long-term changes in the hearing abilities of patients. This formula is known as acclimatization. Another change that takes place following hearing aid use in adults is that the amount off gain preferred by the patient could increase gradually over time. Since hearing losses usually occur gradually, the patient's auditory processing system becomes used to the lower levels of excitation that an impaired cochlea passes on to the CNS. when hearing aids are used, then there is a sudden increase in the output from the cochlea possibly providing greater loudness than the patient is willing to accept. Over the passage of time, the auditory processing system readjusts to the increased cochlear output and hence the patient prefers slightly more amplification, thereby enabling greater speech intelligibility in environments with low speech levels. This aspect of acclimatization is known as adaptation to gain.

Greater the hearing loss, the greater is the increase in loudness when hearing aids are used and greater will be the adaptation to gain over the first few years of hearing aid use. For persons with mild hearing loss the change in loudness produced by hearing aids is so small that there is no measurable adaptation to gain. For persons with severe hearing loss who use hearing aids, it can be inferred that the preferred gain is likely to increase by nearly 10 dB over the next three years of aid use.

Preferred loudness

The prescription procedures for non-linear hearing aids involve amplifying sounds so that they are as loud as, or no louder than for a normal hearing person listening to the same sound. The fact that hearing impaired persons would like to perceive sounds with normal loudness is just an assumption. When loudness is calculated using a loudness model adjusted to allow for hearing loss, hearing aid wearers seem to prefer less than normal loudness. When hearing aid users are asked to assign loudness categories to sounds while listening at the gains prescribed, they as-

sign loudness ratings higher than normal.

Dead regions

The responsibility for sending an electrical representation of sound to the brain stem falls on the inner hair cells of the cochlea. When in a particular region of the cochlea there are no functioning Inner hair cells and no auditory nerves to which they connect, that part of the cochlea is known as the dead region. This would result in a range of frequencies for which there is nor representation within the cochlea. If this frequency is amplified then it is quite possible that they could well be detected by other parts of the cochlea, but the message that reaches the brain stem from these frequencies could be confusing. Intriguingly, persons have better frequency discrimination for frequencies just inside dead regions than for slightly lower or higher frequencies. This is possible because the neurons in the auditory cortex that would normally respond to the dead portion of the cochlea would suddenly find that they have nothing to do and start connecting to the closest nerve fibers that are conveying signals. Despite the presence of copious brain power to analyze the signals spilling over into the functioning frequency regions adjacent to the dead regions, amplification within a dead region contributes less to speech intelligibility than when there are functioning inner hair cells.

Choosing not to amplify too far into a dead region can simplify the fitting by avoiding problems with feedback oscillation and in some cases could slightly improve speech intelligibility. On an average people with dead regions have poorer speech reception thresholds in noise than those who don't, even when the differences in their audiograms are relatively minor.

Regions of the cochlea can be considered to be effectively dead even when inner hair cells have some limited function. If the inner hair cells within a region required a basilar membrane vibration of a high magnitude that sounds at their characteristic frequency are easily detected at some other areas in the cochlea, then that part of cochlea is considered to be effectively dead. Inner hair cells in that region could produce neural responses if the input is amplified enough, but neurons elsewhere in the cochlea will always bee producing stronger neural firings to the same stimuli. These dead regions can be detected by the use of psychoacoustic tuning curves / or the TEN test.

Psychoacoustic tuning curves

This is created by finding thee softest level of narrow-band sounds at various frequencies that just masks a pure tone presented at a small sensation level, typically about 10 dB above its threshold in quiet. If the cochlea has functioning hair cells in the region tuned to the pure tone, then the masker that most easily masks the pure tone will have a centre frequency the same as that of the pure tone. It that part of the cochlea has a dead region, then the person would be hearing the pure tone based on the neural signals created in some other part of the cochlea. This form of neural stimulation is also referred to as off-frequency listening or off-place listening.

TEN test

If a broadband masking sound is presented at the same time as the pure tone signal, then the threshold of the pure tone will be raised whenever the power of the masking sound in the frequency region immediately surrounding the

signal is itself above the threshold in quiet. If the pure tone is being detected at its normal place, then the masked threshold will be similar to the power of the broadband noise falling within the auditory filter surrounding that frequency. Noise that produces equal masked thresholds at all frequencies for people with normal hearing is termed as threshold equalizing noise (TEN). The minimum amount by which the masked threshold has to exceed the threshold in quiet and the minimum amount by which the masked threshold has to exceed the power of the masker within each frequency region before a dead region can be diagnosed are both 10 dB.

A clever modification of this test produces equal masking in dB HL, rather than equal masking in dB SPL. This means that the test can be applied by just re-measuring the audiogram in the presence of TEN noise, and comparing the new masked thresholds to the original audiogram measured in quiet. A step size of 2 dB is recommended for both audiograms. The newer version of the noise also has a more restricted bandwidth and has noise with a low crest factor to decrease the likelihood of it producing loudness discomfort during the test.

Usefulness of testing for dead regions in cochlea depends on what prescription would have been given in the absence of the results of this test. Assuming one made the assumption that the benefit of achieving audibility within each frequency region did not depend on the degree of hearing loss within each region - one would then prescribe a large gain at every frequency for which the loss was severe. If some frequencies were within a dead region, a significant error would have been made - the amplification at best would be wasted, and at worst would cause feedback oscillation and reduce speech intelligibility.

If prescription is based on the assumption that the greater the loss in a frequency region, the smaller the contribution that region will make to intelligibility, no matter how much audibility is achieved. Learning that the frequency region has not functioning inner hair cells may not significantly change the prescription.

There is no 1:1 correlation between the degree of hearing loss and the presence of dead regions, it is certainly true that the likelihood of a dead region greatly increases as the degree of loss increases. Even if the prescriptive procedure has not explicitly allowed for the reduced effectiveness of frequency regions with severe hearing loss, experienced audiologists are likely to over-rule the prescription and decide that some frequency regions have too much loss to be aidable. On the basis of audiogram alone, experienced clinicians typically decide that hearing loss greater than about 90 dB HL is not aidable. Given the potential measurement problems, clinical time involved, and lack of certainty about how to use the result, most prescription formula do not require measurement of dead regions as a mandatory input to the prescription formula. The CAM 2 prescription is however intended to be applied at frequencies where there are no dead regions. It should be stressed that the impact of dead regions on requirement of amplification is plausible and could find its way into prescription formulae to a greater extent in the future.

Severe hearing loss,s effective audibility and high frequency amplification

As frequency progressively increases beyond 2 kHz, several factors play a role in making things difficult for the patient and the treating clinician. When the intensity of speech weakens hearing

loss also increases. Both these factors could mean that more gain is needed to achieve audibility. Speech information exists for frequencies of up to approximately 10 kHz, the amount of information per 1/3 octave band decreases as frequency increases. This decrease is compounded by the patient's decreasing ability to use the information, even if it is made audible if the hearing loss increases with frequency.

Optimal upper frequency limits are difficult to estimate accurately because:

1. The greater the high-frequency hearing loss, the smaller the value of an extended high-frequency response.

2. The greater the unique information present in the individual talker's voice at high frequencies, the wider the optimal bandwidth will be. Audibility above 4 kHz appears to be more important for the perception of fricatives spoken by females than by males. Inadequate audibility of very high frequency sounds is likely the cause of late development of fricative production in hearing-impaired infants.

3. The shape of the noise spectrum relative to the speech spectrum will affect the value of a higher frequency limit. When Signal to noise ratio increases as frequency increases, there will be more value in extending the frequency limit than when SNR decreases as frequency increases. High-frequency speech cues become more important when low-frequency cues are not available, such as when they are masked by noise.

4. Spatial separation of speech and noise generally improves speech intelligibility, and high frequency cues contribute to this improvement. Wider bandwidths are hence more important when the target sound is spatially separated from competing sounds.

Excessive levels of sound presentation would reduce its intelligibility, and in particular excessive high-frequency stimulation may mask information at lower frequencies (downward spread of masking). A hearing impaired person will show no benefit for a wide bandwidth if the gain is insufficient to provide any audibility across that extended bandwidth.

Higher bandwidths tend to be preferred by persons with flat losses, and lower bandwidths by persons with steeply sloping losses. Those with steeply sloping losses are most likely to have had high-frequency dead regions and hence are not benefited from high frequency amplification. The decreased ability of the impaired ear to extract information form a signal even when it is audible is referred to as hearing loss desensitization.

The need for a small but positive sensation level over the high-frequency range to maximize speech intelligibility brings with it some challenges. To retain a small sensation level as the input level changes over a wide range requires a very high compression ratio. Slow acting compression in the hearing aid would help.

Prescribing compression thresholds

Compression threshold is the input level above which the gain of the hearing aid starts reducing as the input level increases. Majority hearing aids have multiple channels of compression, making it difficult while describing compression thresholds. Compression threshold can also be expressed as the overall level of a broadband signal with some particular spectral shape at which

compression begins in some or all channels, or it can be expressed as the level of the signal within any one channel at which that channel enters compression. The greater the number of channels, the narrower in frequency each channel is, and the smaller the within-channel compression threshold will be relative to the overall, broadband compression threshold.

If loudness is to be normalized completely by a hearing aid, compression is needed for input levels from the threshold of normal hearing upwards. Compression threshold must be around 5-10 dB SPL. The gain for low-level sounds will equal the hearing loss. This requires the mold/shell to be much more tightly sealed than would be necessary for a linear hearing aid.

NAL-NL prescription on broadband compression is based on compression threshold of 52 dB SPL. This is so because of the fact that it is near the bottom of the range of speech levels normally encountered by people. When used with slow-acting compression however, values lower than this is almost optimal. Extra emphasis should be given to evaluating the suitability of compression threshold after the aid wearer has had a chance to try the hearing aid in his/her usual environments. It has been shown that less compression is appropriate for people with severe and profound loss than for persons with moderate loss.

Prescription should achieve three main goals:

1. To give the best speech intelligibility possible for that patient

2. To provide overall loudness that is acceptable to the patient

3. To give a tonal quality preferred by the patient.

Speech intelligibility is predominantly affected by the amount of signal audible above both hearing thresholds and background noise. In many situations it is the background noise that determines the SNR across most or all of the frequency range. In these situations, for speech inputs at typical conversational levels, the gain-frequency response can be varied considerably without affecting audibility at each frequency, and hence without affecting speech intelligibility. Most prescription procedures attempt to amplify speech with typical input levels (around 60-65 dB SPL) to give a comfortable overall loudness.

Gain, Frequency Response, and Input-output functions for Non-linear amplification

Non-linear prescription can be viewed as specifying the gain-frequency response for several input levels. Typically both the average gain and shape of the frequency response will vary with input level. The prescription can be viewed as specifying an input-output curve for several frequencies. It is necessary to specify the I-O curve for at least as many frequencies as there are channels in a multichannel hearing aid.

Filters are used to form the individual channels in a multichannel aid, and can also be used to shape the frequency response within each of these channels. A filter providing different amounts of gain at different frequencies and levels can shape the signal in much the same way, but without forming separate channels and recombining their outputs.

LGOB (Loudness Growth in half-octave Bands)

The concept of using non-linear amplification to

restore normal loudness perception has been there for nearly 35 years. In this procedure, the hearing impaired patient categorizes the loudness of narrow bands of noise using a seven-point loudness scale.

1 - Not audible

2 - Very soft

3 - Soft

4 - OK

5 - Loud

6 - Very Loud

7 - Too loud

IHAFF / Contour

During 1990's a group of audiologists noted that there was a urgent need for a practical procedure that could be applied to any hearing aid with adjustable wide dynamic range compression. The group was called the Independent Hearing Aid fitting Forum and the prescription process they devised used loudness scaling to normalize loudness at each frequency. This loudness scaling procedure is known as the contour test.

A software called VIOLA (visual input/output locator algorithm) simplifies the task of calculating the input-output curve, based on the contour test results.

ScalAdapt

For all the above described normalization procedures hearing aid prescription is a three step process:

1. The loudness scale for the patient is measured

2. At each level, the gain needed to normalize loudness is calculated

3. The hearing aid is adjusted to match the gain target

ScalAdapt is actually a one-step combination of all the above stated three steps. The hearing aid is pre-adjusted using an established threshold-based procedure. Loudness scaling, using a 11-point scale is then performed while the patient is wearing the hearing aid. Instead of finding the loudness that corresponds to each input level, the clinician adaptively adjusts some characteristic of the hearing aid until the patient gives a desired loudness rating. This desired rating is the rating that would be given by a normal-hearing person listening unaided to an input of that level.

For example, if a normal-hearing person would rate a sound of 60 dB SPL at a particular frequency as comfortable, then the gain of the hearing aid is adjusted until the hearing-impaired person also rates a 60 dB SPL sound at that frequency as comfortable. The hearing aid parameters are adjusted adaptively.

FIG6

This procedure specifies how much gain is required to normalize loudness, at least for medium and high level input signals. This procedure is not based on individual measures of loudness, but it uses loudness data averaged across a large number of people with similar degrees of threshold loss.

This procedure gets its name from Figure 6 of the article in which the underlying data were first outlined. Gain is directly prescribed for each of the input levels 40, 65, and 95 dB SPL and is inferred for other levels by interpolation.

For low level (40dB SPL) input signals, the gain is prescribed on the basis that people with mild / moderate hearing loss should have aided thresholds 20 dB above normal hearing threshold. In most circumstances, it is not worth providing more gain than this, as background noise will prevent very soft sounds from being perceived no matter how much gain is prescribed. Except for the first 20 dB of hearing loss, every additional decibel of hearing threshold loss is therefore compensated by an extra decibel of gain. This rule is relaxed to a half-gain rule once the unaided threshold exceeds 60 dB HL because otherwise the high gains that result are likely to cause feedback oscillation.

DSL[i/o] and DSLm[i/o]

The desired sensation level (DSL) prescription for non-linear hearing aids has undergone various modifications in line with experience and new data since the first non-linear version appeared in 1995. The first non-linear version is called DSL[i/o], actually comprised of two alternative procedures each with its own underlying rationale. One procedure was called DSL[i/o] linear, where linear means that the I-O curve is a straight line over a wide range of input levels. That is, the compression ratio is constant within the wide dynamic range compression region. This should not be confused with linear amplification. This procedure uses a compression ratio large enough to fit an extended dynamic range at each frequency into the dynamic range of the hearing impaired person at the same frequency. This extended dynamic range is equal to the range from a normal-hearing person's threshold up to the hearing impaired person's uncomfortable level.

DSLm[i/o] in this prescription, m stands for multistage, comprising limiting at the highest levels, WDRC (wide dynamic range compression) across he mid levels, linear amplification below compression threshold.

NAL-NL1 and NAL-NL2

These methods do not attempt to restore normal loudness at each frequency. The rationale behind this prescription protocol is to maximize speech intelligibility, subject to the overall loudness of speech at any level being nor more than that perceived by a normal hearing person. The only inputs required by both these models are hearing thresholds, and the speech spectrum levels input to the ear after amplification.

For speech input at any level, gain at each frequency was systematically varied within a high-speed computer until the calculated speech intelligibility was maximized, but without the calculated loudness exceeding the loudness calculated for normal-hearing persons listening to speech at the same level.

Advantages of choosing threshold-based prescription formula:

1. It is fast

2. Can be used in all patients

3. Loudness can be partially predicted from thresholds

4. There is no evidence that loudness is critically important as opposed to audibility

5. Loudness normalization for narrow band test stimuli in a test booth may not achieve normal loudness for broadband stimuli in the real world

6. Normal loudness is ill defined because it varies considerably across people and across measurement techniques

Advantages of loudness based prescription formula:

1. Individuals with the same audiogram could perceive different loudness for the same sound

2. Achieving normal loudness is a worthwhile goal, as well as a means to achieving audibility and intelligibility

3. Accurate prescription of overall loudness is important for hearing aids with no volume control

Prescription formula for conductive hearing loss and Mixed hearing loss

All the above stated prescription is applicable for sensorineural hearing loss. A conductive loss or a conductive component in a mixed loss comprises of a frequency-dependent attenuation of sound in the middle ear. In pure conductive hearing losses, hearing threshold, MCL (Most comfortable level) and LDL (Loudness discomfort level) are all elevated by the same amount and this elevation equals the amount of attenuation occurring in the middle ear. In mixed hearing losses, it can be assumed that the conductive component also causes all three quantities to increase approximately by the same amount. The size of the conductive component at each frequency is inferred from the size of the air-bone gap on the audiogram.

Taking all these factors into consideration it is ideal to prescribe a person with conductive hearing loss, insertion gain at each frequency should just equal the conductive loss at that frequency. This compensation would cause a normal input into the cochlea, which itself is normal.

In patients with mixed hearing loss when fitted with a hearing aid, the average gain needed equals half the total loss plus one quarter of the conductive component.

One aspect that needs to be taken into consideration is that acoustic reflex causes low-frequency sounds entering normal ear at high levels to be attenuated by middle ear muscle contraction. This reflex is absent in the case of conductive impairment. The hearing loss at low frequencies for high level sounds is a little less than that of low level sounds, consequently for low frequency high level sounds, the gain required to provide a normal input to the cochlea is less than the elevation in hearing thresholds.

Synopsis of high-frequency amplification

1. High-frequency information in speech is found around 10 kHz, particularly for fricatives and in female speech

2. The greater the hearing loss, the smaller the amount of information that persons can extract from audible speech. The deficit relative to normal hearers increases as sensation level increases. This deficit applies equally at all frequencies, but will have adverse effects much more often in the high frequencies because hearing loss is often the largest in the high frequencies.

3. Excessive high-frequency amplification sensation level can decrease speech intelligibility, but most often just cause additional loudness and poor speech quality without impacting speech intelligi-

bility.

4. Excessive high-frequency amplification arises from excessive gain and hence sensation level, rather than excessive bandwidth. The bandwidth over which some audibility should be provided should always be as wide as possible.

5. It is ideal to use a prescription procedure that attempts to provide a wide bandwidth and sensation levels that reflect both the importance of different frequency regions and the impact of hearing loss on the ability to extract information.

Excessive amplification and aggravation of hearing loss

Hearing aids amplify sound. They hence have the potential to cause a noise-induced hearing loss to the user who already has a hearing loss. Aggravation of hearing loss caused by usage of hearing aids depends on the following factors:

1. A person's susceptibility to noise-induced hearing loss partly depends on how much loss the person already has.

2. Noise exposure that causes a certain permanent threshold shift to someone with normal-hearing will cause much less threshold shift in someone with a severe loss.

3. Daily noise dose is another factor that affects noise-induced loss in the aid user. This value depends on the levels at the output of the aid and the amount of time that these levels are maintained. The input level fluctuates with time, and so does the output level too. It is hence impossible to describe the output level as a single representative number.

Output level of hearing aid depends on three factors:

1. Greater the gain, the greater will be the output level.

2. Greater the level of sound at the input to the aid, the greater will be the output level.

3. Both the above statements are true only when the output is less than the maximum output limit of the aid.

Output sound pressure level 90 : This is the output saturation sound level (SSPL) to a 90 dB sound input and is measured over a frequency range. This is also known as SSPL90.

Noise dose is determined by the combination of the input level and gain, rather than by the OSPL90 value. This fact is important as it is often incorrectly assumed that the safety of the hearing aid is determined solely by its OSPL 90 value.

Avoiding hearing aid induced hearing loss

1. Do not prescribe more gain or OSPL90 than is necessary for optimal intelligibility. This is very important for children too young to operate their remote control, or anyone fitted with a hearing aid that has no volume control.

2. Patient should be advised to avoid prolonged exposure to high noise levels.

3. Non-linear hearing aids in which the average gain decreases as the input level rises from typical input levels to high input levels should be prescribed.

4. Hearing threshold levels should be monitored over time

5. When doubt exists, check should be made for temporary threshold shift by measuring hearing thresholds after 24 hours without a hearing aid in the test ear and then after 8 hrs of hearing aid use.

Hearing Aid Selection & Verification

The first decision that needs to be made by the clinician and the patient is to select the type of hearing aid from the following choices:

1. CIC

2. ITC

3. ITE

4. BTE-RITE

5. BTE-RITA

For each of these styles there are advantages relating to ease of insertion, ease of control manipulation, visibility, amount of gain, sensitivity to wind noise, directivity, reliability, telephone compatibility, ease of cleaning, avoidance of occlusion and feedback, ability to assess and fit in the same appointment and cost. The importance given to each of these factors will vary between patients.

Need for specific features, such as volume control, telecoil and switch, a direct audio input and a directional microphone must be determined on an individual basis. These needs will also influence the style of hearing aid selected. BTE's have more advantages when compared to other styles in a majority of patients.

Signal processing options appropriate to the needs of the patient should be selected. Compression limiting is a more appropriate form of limiting than peak clipping if it can provide a high enough maximum output. In addition to compression limiting feature a low compression ratio which is active over a wide range of input levels is appropriate for majority patients. This low-ratio compression could provide advantages whether it is single or multichannel, or whether it is fast or slow acting. Multichannel compression could provide greater speech intelligibility and comfort for patients with moderately or steeply sloping hearing loss and the multichannel structure facilitates other features such as adaptive noise suppression and feedback suppression. Adaptive noise reduction feature is very useful for patients who wear their aids in a range of environments and who also require amplification across a wide range of frequencies.

Feed back cancellation would be beneficial for patients with a severe or profound hearing loss, patients with a severe loss in the high frequencies but near-normal hearing in the low frequencies.

Frequency lowering could be advantageous in some patients though it is currently not possible to predict which of them would benefit.

Hearing aid fitting software provides a first approximation to the prescribed gain-frequency response target. The software should allow for the acoustic configuration of the earmold shell or dome fitting. The approximation can be made even more accurate by incorporating the individual patient's real-ear to coupler difference (RECD) in the prescription.

Selecting Hearing aid style: (CIC, ITC, ITE, BTE, Spectacle aid, or body aid)

There are many factors that are considered before selecting a hearing aid style. The spectacle and body worn hearing aids are not popular and is rarely being used these days. In rare instances where spectacle aids are used, they are mostly implemented by attaching a spectacle adapter to a BTE hearing aid.

The ease of management generally affects the success of a hearing aid fitting. The older the patient, the more important the ease of management becomes.

Ease of insertion & removal:

ITE, ITC and CIC hearing aids are easy to insert and remove because they come in a single pack-

Factor	CIC	ITC	ITE	BTE/Mold	BTE/dome/ RITA	BTE/dome/ RITE	Spectacle	Body
Ease of insertion/removal	✓✓	✓✓	✓		✓	✓	✓	✓
Ease of manipulation of controls		✓	✓✓	✓✓✓	✓✓	✓✓	✓✓✓	✓✓✓
Invisibility	✓✓✓	✓✓	✓	✓	✓✓	✓✓		
High gain & max output			✓	✓✓		✓	✓✓	✓✓✓
Bandwidth & freq shape	✓✓✓	✓✓✓	✓✓✓	✓✓	✓✓	✓✓✓	✓✓	
Insensitivity to wind noise	✓✓✓	✓✓	✓✓					
Directivity (omni Directional mic)	✓✓✓	✓✓	✓					
Directivity (directional mic)		✓	✓✓✓	✓✓✓	✓✓	✓✓	✓✓✓	
Reliability				✓✓✓	✓✓✓		✓✓✓	✓✓✓
Telephone compatibility	✓✓	✓	✓	✓✓✓	✓✓	✓✓	✓✓✓	✓
Ease of cleaning				✓✓✓	✓✓✓		✓✓✓	✓✓✓
Avoidance of occlusion		✓	✓✓	✓✓	✓✓✓	✓✓✓	✓✓	✓✓
Avoidance of feedback			✓	✓✓✓			✓✓✓	✓✓✓
Same day fit					✓✓✓	✓✓✓		
Cost	✓	✓	✓✓	✓✓	✓✓	✓✓	✓✓	✓✓

Table showing relative advantages of different types of hearing aids

age and they do not interfere with spectacles. BTE hearing aids with no helix lock on the mold may be easier to insert than ITE hearing aids with a helix-lock. For some users, full concha ITE hearing aids are harder to insert than ITC and CIC aids because of the difficulty faced in inserting the helix lock. CIC's with removal strings are relatively easy to insert and remove. Deep seated long-wear CIC's do not present any insertion and removal difficulties because the clinician takes full responsibility for insertion and removal which is needed only when the battery has drained out.

Ease of on-aid user control manipulation

It could be difficult for the aid wearer to manipulate a control on a CIC hearing aid while it is in the ear, especially so if it is deeply inserted. Gain adjustment becomes easier if an extended flexible shaft is attached to the volume control, but this makes the aid more obviously visible doing away with the cosmetic advantages of the CIC aids. The controls on body worn aids, spectacle aids, and larger BTE aids are easier to operate because they are larger and are easily accessible because of their location. Add on caps can often by applied to increase the height and ease of use of a volume control for an ITE or ITC device, but at the expense of their appearance. This may not be an issue if automatic gain control via compression is adequate for the patient, or if a remote control is available and acceptable for the patient.

Invisibility

CIC hearing aids are very discrete as they have very low visibility. Deeply seated CIC hearing aids have total invisibility as they reside deep within the external auditory canal. Small BTEs with thin tubes or connecting wires could be almost invisible, especially in patients with hair that can be used to hide the aid. RITE-style (Receiver in the ear) BTEs can have smaller cases than RITA-style BTEs (Receiver in the Aid) because the receiver need not fit within the case as it is placed within the ear canal. Manufactures make RITA-style hearing aids still more smaller by eliminating telecoils, user switches and direct audio input. The use of size 10 batteries still make the size smaller. Some manufacturers disable low-frequency channels of open-fit hearing aids to decrease battery current to facilitate use of small batteries while retaining a reasonable battery life.

High gain and maximum output

The further the hearing aid microphone is from the ear canal entrance the greater will be the gain without feedback. The larger the receiver and battery, and hence larger the hearing aid the greater will be the OSPL 90 value, particularly in low frequencies. Open fitting BTEs combined with effective feedback cancellation can achieve effective feedback cancellation and achieve high-frequency insertion gain of around 30 dB. This amount of gain would match the prescription for high-frequency losses of up to about 60 dB HL.

The dome-style canal pieces which are non-customized are unsuitable for achieving significant low-frequency gain. Open-dome will not allow any low-frequency gain to be achieved at 250 Hz and greatly restrict the gain achievable at 500 Hz. Some low-frequency gain is achievable with closed domes, especially those with double flanges, but unfortunately the amount of leakage around a closed dome is highly variable, so they are more or less satisfactory than a custom mold if low-frequency gain is required. Low-frequency gain is required for low-frequency losses of 25 dB or greater if one wishes to match the prescription for all input levels above 50 dB SPL, and for losses of 30 dB or greater to match the prescription at 65 dB SPL and above.

Image 1.73 showing the types of Behind the ear hearing aids

Bandwidth and frequency response shape

Within BTEs, the lack of tubing resonances in RITE hearing aids enables slightly higher gain and OSPL90 in the very high frequencies and a smoother response shape across the mid and high frequencies compared to the RITA style, especially if it has no damping. Ofcourse, electronic filters can be used to remove the peaks and dips caused by the tubing, but this always comes at a cost of reduced OSPL90. The sound quality advantages arising from the inherently smoother response of the RITE style may be responsible for the subjective preference for the RITE style. This has to be traded off against the greater reliability of the RITA style arising from the receiver being not within the ear canal.

Thin-tube BTEs have slightly poorer high-frequency response than BTEs delivering their sound with wider tubing, markedly so if the latter employs a high frequency horn.

Insensitivity to wind noise

Most wind noise is generated by turbulence created by head and the pinna. CIC hearing aids pick up less wind noise than other types of hearing aids because the microphone is further from the turbulence producing parts of the pinna and head. CIC microphones are also protected from the direct flow of wind. BTE and spectacle hearing aids are affected by wind noise. Hearing aids with directional microphones are extra sensitive to wind noise.

Many hearing aids automatically provide a low-frequency cut when they detect wind noise, this cut also cuts the low frequency content of speech.

Directivity

BTE, ITE and larger ITC hearing aids are the only styles of hearing aids big enough to contain a directional microphone and are thus able to suppress sounds coming from the side and rear of the head. Spectacle hearing aids equipped with multi-microphone array, and true binaural signal processing

have potentially the best performance, but these microphone arrays have so far been limited only to research studies and are not commercially available.

If only omni-directional microphones are considered CIC aids have the best directivity followed closely by ITC aids, since these aids make the greatest u se of the sound collecting and sound attenuating properties of the head, pinna and concha. Micro BTEs sometimes sit so fat behind the pinna that the directional microphone is less effective. Micro BTEs are often fitted as open-canal devices, which dramatically reduces directivity when averaged across frequency.

Reliability

The greatest threat to reliability are moisture and cerumen. Hearing aids in which the receiver is located in the ear canal are the least reliable, because cerumen and moisture limits the life of the receiver. Wax guards are useful for reducing wax ingress. The most unreliable parts are those that involve movement / electrical contact between moving surfaces such as switches, volume controls and battery contacts. Nanocoating and waterproofing are known to improve the reliability by reducing or eliminating moisture ingress.

Telephone compatibility

Hearing aids can pick up acoustic signals or magnetic signals coming from telephone handsets. For non-micro BTE and spectacle aids, telecoil mode can easily be selected and used. For body worn aids the body worn unit should be held near the phone handset, and this really complicates its usage. In ITE and ITC aids a telecoil selector switch makes the faceplate more crowded and increases the difficulty of operating the controls, particularly if the hearing aid has a volume control. This can partially be overcome if the hearing aid is provided with a remote control, or if the hearing aid automatically selects telecoil input when it encounters a strong magnetic signal.

In many hearing aids, the telephone can simply be placed over the ear so that hearing aid amplifies the acoustic signal. This frees the patient from the task of selecting telecoil mode. This is possible only if the proximity of the telephone does not cause feedback oscillations in the hearing aid. Feedback cancellation processing in the hearing aid / acoustic damping material placed over the telephone receiver helps to avoid this problem.

The compatibility with wireless receiver that receives signals from a streaming interface device, which in turn receives blue-tooth signals from the telephone is rather excellent with the current generation hearing aids.

Stethoscope compatibility

Hearing impaired medical practitioner could use a stethoscope while wearing a CIC without causing feedback oscillation, provided the CIC does not have too much gain / leakage of sound past the shell. Possible alternate solutions to this scenario could be:

1. Remove the hearing aid, and use amplified stethoscope.

2. Use of an amplified stethoscope for which the output device is a pair of supra-aural or circum-aural earphones, which are placed over the ears and hearing aids together.

3. Couple the output of an amplified stethoscope to the direct audio input or telecoil input of the

hearing aid or to a streaming interface that wirelessly transmits to the hearing aids.

Avoidance of occlusion and feedback

Patients with near-normal low-frequency hearing combined with a severe high-frequency loss are difficult to fit accurately despite the fact that they have a very common hearing loss configuration. The low-frequency thresholds require a large vent or an open fitting to minimize occlusion but the gain required at high frequencies to fully meet prescription targets may cause feedback oscillation even if feedback cancellation technology is applied. Studies reveal that feedback can be reduced by increasing the distance between the vent outlet to the microphone inlet. This is easy to achieve in BTE hearing aids than in ITE, ITC or CIC hearing aids. Deeply seated CIC hearing aids even with soft tips are likely to result in discomfort to the patient causing low levels of satisfaction.

Same-day assess and fit

BTE hearing aids do not require a custom ear mold and can be provided to the patient in the same appointment during which the needs of the patient is assessed. If the patient is willing then hearing aid can be fitted immediately without delay in the same sitting. Studies reveal that same-day assessment and fitting is possible for nearly 80% of patients. Feedback and oscillation problems were reported by nearly a third of them which was corrected during later sittings. The proportion of patients who needed non-custom earmolds decreased as both hearing thresholds and age increased.

Cost

CICs cost more than other hearing aid styles because miniaturization costs more. BTEs cost less to procedure than custom hearing aids.

Battery size

A small hearing aid cannot accommodate a large battery. As the size of the battery decreases, handling difficulties increase and more features cannot be added to the aid due to the fear of excessive drainage of the battery.

Points to be considered in selecting hearing aids to minimize management problems:

1. Hearing aid with a wide dynamic range compression with no volume control is chosen.

2. Largest hearing aid style with a large battery can be prescribed provided it is cosmetically acceptable. Half concha ITE should be chosen instead of BTE.

3. In patients with good mental capabilities but with poor physical manipulation capabilities a hearing aid with a remote control.

4. For patients with poor vision, all controls and the battery compartment opening point should be located tactually.

Compromises is often necessary while prescribing hearing aids. If the patient places too much importance on invisibility of the hearing aid then a micro BTE with an open dome canal piece will meet these aims. It is always wise to prioritize factors that enable the patient to accept the hearing aids and get some benefit from it.

It should be noted that the provision of less than optimal high-frequency gain to achieve physical comfort and good own-voice quality is a compro-

mise.

Compromises may not always be in the direction of prioritizing comfort and appearance over performance. Patients with a mild flat or gently sloping hearing loss will appreciate the absence of occlusion effect provided by an open fitting, because their greatest need being improved speech intelligibility in noise is satisfied by this setup. Intelligibility in noise will be maximized by a fitting that is directional over the widest possible frequency range, which is not facilitated by an open fitting.

As far as the spectacle based hearing aids the ophthalmologist and audiologist will have to coordinate their activities to ensure that the spectacle adapter on the hearing aid matches the spectacle frame.

Selection of hearing aid features

The following features should be considered before taking a final decision on the hearing aid model.

Volume control - All varieties of compression decrease the need for a volume control, although not necessarily to the same degree. Many patients feeel happy not to have to manipulate a volume control. Many of them ofcourse will not need one if the hearing aid has a wide dynamic range compression (WDRC) with an adequately high compression ratio and adequately low compression threshold. There are of course three categories of patients who would benefit by the presence of volume control. They are:

1. Those for whom the WDRC does not achieve an acceptable loudness in some situations.

2. Those who psychologically strongly desire to control their hearing aids.

3. Those experienced users who are used to volume control.

Volume controls could cause problems for some patients if they are accidentally moved. There is no effective way to predict which patients are likely to need a volume control. It is always hence safer to order a manual control that can be electronically locked.

Another way in which benefits of volume control can be provided to the patient is actually fitting the one with a push-button that can be used to select from multiple preset memories. Each push of the button would advance the hearing aid to its next memory. If the aid wearer is comfortable using a remote control then there is very little need for a volume control within the hearing aid.

Telecoil

This is essential for anyone with a severe / profound hearing loss. It is also useful in persons with moderate hearing loss to keep using their telephones. Persons with mild hearing loss usually can cope up well with the telephone without their hearing aid. Persons with all degrees of hearing loss would appreciate the reduction in noise and reverberation that telecoils offer when used in conjunction with a room loop.

Main disadvantage of a telecoil is the increase in size of the aid needed to fit in the telecoil. If the presence of the telecoil makes it necessary to add a program switch, then a second disadvantage crops up due to the increased crowding that occurs in ITE / ITC faceplate or micro BTE case. This crowding could make it difficult for the patient to find and operate the correct control. These disadvantages should be weighed against the substantial

advantages offered by the hearing aid. Commonly, persons with severe to profound hearing losses have telecoils while persons with mild losses don't need it. Automatic telecoil selection, especially by hearing aids that communicate across the head wirelessly so that they make good decisions about the proximity of a phone, enables a telecoil to be included without the need of a selector switch.

Direct audio input / wireless input

Direct audio input is useful in the following scenario:

1. Persons who use a wireless transmission system is electrically coupled to their hearing aids to improve SNR. Adults as well as children can benefit immensely from a wireless system. The sensitivity of the FM system and hearing aid should be carefully adjusted by the clinician so that the combination gives the high SNR.

2. Persons who use a hand-held directional microphone connected to the hearing aid via a cable. Commonly persons with severe / profound hearing loss usually choose to use these devices. The increase in signal-to-noise ratio can be substantial. These microphones can provide directivity superior to that of head-worn microphones and often can be held closer to the source.

3. Person who watch Television in a noisy environment the unit that contains a microphone that can be placed close to the TV or plugged directly to the TV out can make it very useful in increasing the SNR.

Directional microphones

Directional microphones can offer a substantial improvement in signal-to-noise ratio. Hearing aids can be ordered with directional microphones permanently selected, but most hearing aids with directional microphones can be switched or automatically switch between directional and omni-directional modes. The only reason for not choosing a switchable directional microphone is if the patient wants a low-visibility custom hearing aid (CIC or ITC). It is not possible to fit an effective directional microphone into these hearing aids. Currently some larger ITC hearing aids have directional microphones with limited directivity.

Disadvantages of directional microphones

1. Directional microphones are even more prone than omni-directional microphones to wind noise. They can be a problem in persons who spend a lot of time outdoors.

2. In certain circumstances it is not possible for the patient to always look at the sound source. This happens while driving a car and listening to passengers. In these situations speech and environmental sounds may be clearer and audible when omni-directional microphone is used.

Two aspects of the patient's hearing loss may limit the range of frequencies over which directivity can be achieved and hence could limit the effectiveness of the directional microphone:

1. Patient with a severe-to-profound hearing loss may require a low to mid frequency response considerably flatter than can be provided by a hearing aid with a directional microphone. Achieving a flat response is rarely a problem, but greater gain and lower internal noise in the low frequencies can always be achieved with an omni-directional microphone than with directional microphone. In these patients directivity can be limited to the high frequencies.

2. Patients could require amplification only over a restricted frequency range (above 1500 Hz). The hearing aid will be directional only over the frequency range where amplified sound dominates over vent-transmitted sound.

It has been suggested that a speech-in-noise test could be used to determine the SNR deficit before determining whether a patient needs directional microphones. Every person with a hearing loss has trouble understanding speech when the SNR is sufficiently poor.

Compression limiting versus peak clipping

Peak clipping should be chosen in preference to compression limiting only for:

1. Patients with a profound hearing loss who need the greatest possible OSPL90. If patients prefer the volume control to be turned to its highest setting, it is likely that they would benefit from more gain. Greater maximum output, especially for speech signals is possible with peak clipping.

2. Patients who have to be fitted with a larger hearing aid to achieve the required OSPL90 in a compression limiting aid, but who do not want to wear a larger hearing aid.

Wide dynamic range compression (WDRC)

There is evidence to suggest that WDRC with a low compression ratio should be available in all hearing aids. A small proportion of patients may not gain any advantage from WDRC relative to linear amplification. There is no reliable way to predict who these patients could be. WDRC is likely to be more advantageous for high cognition clients who need their hearing aids in a wide variety of communication situation who have sloping hearing losses. It is safe to initially select some form of WDRC for everyone. For persons with profound hearing loss, relatively high compression thresholds / low compression ratios may be necessary.

Multichannel compression

Multichannel compression will provide additional benefit for patients with a moderately or steeply sloping hearing loss, because a different degree of compression can be used in each channel. It is hence ideal to use multichannel compression for any patient whose 2 kHz treshold exceeds the 500 Hz threshold by more than 25 dB. These patients derive more benefit from TILL response hearing aid (Treble increases at low levels). Patients suffering from flat hearing loss will not benefit from single channel compression hearing aids. Multichannel compression is the most common way to implement adaptive noise suppression and adaptive microphone directivity.

Fast-Slow acting compression

Advanced multichannel hearing aids currently available in the market have very fast-acting compression, while others have very slow-acting compression. Some hearing aids have combinations of fast and slow acting compression while others have programmable attack and release times. At present there is no way of systematically deciding which type of compressor is best suited for the patient. Indications are that fast acting compression is best suited for subjects with a high level of cognitive functioning as measured by their ability to identify target sequences amongst rapidly changing visual patterns and who frequently need to use their hearing aids in communication situations where the sound level varies rapidly by large amounts. Slow acting compression is likely to benefit those

persons who wear hearing aids in a range of environments that have different mean sound levels or in which sound levels change rather slowly.

Ideally speaking both forms of compression should be available in the hearing aid so that the overall compression ratio can be increased while minimizing the disadvantages associated with a high compression ratio of either type.

Adaptive noise reduction

Amplification schemes in which the gain is automatically reduced in those frequency regions that have the poorest signal-to-noise ratio are most likely to be appreciated by patients who wear hearing aids in a wide variety of noisy environments. The benefits of adaptive noise reduction will be far greater for patients who need amplification for all frequencies than for those who need amplification for only the high frequencies. If the low-frequency gain of a hearing aid is dominated by vent transmitted sounds, the effects of adaptive noise reduction will be confined to higher frequencies which could make the adaptive noise reduction process a little less effective.

Multiple memories

Candidacy issues for multi-memory hearing aids are very much similar to those for adaptive noise reduction. Users are likely to benefit from multi-memory amplification if they require amplification over a wide frequency range, regularly wear hearing aids in acoustically diverse listening environments, and have more severely restricted dynamic range in high frequencies. Both multi-memory amplification and adaptive noise suppression aim to vary the amplification characteristics depending on the acoustic environment and both these devices can achieve useful gain variations only at those frequencies where the gain is greater than 0 dB.

Multiple memories can be used to access telecoil or to change the microphone directionality. Auto switching between memories depending on environment is acceptable to many patients. The environment detectors of course are not perfect and some users find it disconcerting if the program and hence the sound quality changes in the absence of any distinct change in the environment.

Feedback management schemes

This will benefit persons with:

Profound hearing loss.

Good low-frequency hearing combined with poor, but usable high frequency hearing.

Who wear open-canal hearing aids.

Who use telephone on microphone setting.

If hearing aid has a tendency to oscillate / ring then enabling feedback cancellation will be really beneficial. If oscillation or ringing is not a problem, feedback cancellation will cause no adverse effects with possible exception of its effect on musical sounds.

Frequency lowering

It is really unclear for which patients frequency lowering should be enabled. There seems to be little to lose in enabling it for patients for whome typical level speech cannot be made at least partly audible across the frequency range from around 3-6 kHz or in patients with known dead regions.

Trainability

Some users would appreciate the opportunity to train their hearing aid to their personal preferences in their actual listening situations. They need to be provided the opportunity to train their hearing aids according to various acoustic situations.

Selection and adjustment of programmable hearing aids

This involves 12 steps.

Step 1: Entering audiometric data

As a minimum audiometric data which include pure-tone thresholds for each ear to be fitted should be entered into the programming tool. Other data that need to be entered include patient identifying data, discomfort levels, most comfortable levels, loudness scales, speech identification scores, acoustic reflex data and tympanometric data. There will be two modules available in this programming tool. One is the clent module in which the client identification details are entered while the other is the audiometry module where the other desired hearing parameters are entered.

Step 2: Open manufacturer's software

There is no really systemic way to choose a particular brand of hearing aid for a patient. Major manufacturers have comprehensive ranges of hearing aids which could be used by a majority of patients. Factors that could decide the manufacturer are:

History of reliable and timely sales and after sales service.

Discounts applicable.

Step 3: Select a fitting method

Most manufacturers offer the choice of generic prescription procedures like (NAL-NL2, CAM2, FIG6 etc). Some others only offer a proprietary prescription procedure developed by them.

Step 4: Earmold and earshell options and their selections.

Some manufacturer's software will automatically recommend a vent size. Some software isntead requires the clinician to specify the earmold / earshell configuration. Other software makes no allowance for earmold or earshell configuration. It is important to specify an approximate vent size of 2 mm. If it is not specified then the software could make large errors in calculating the coupler gain needed to achieve the real-ear gain target.

Step 5: Select a potential hearing aid.

In majority of software the initial desired specification of the hearing aids should be entered. The exact procedure for this step varies between manufacturers. At one extreme one is required to specify which particular hearing aid style that the user is interested in. The clinician should also indicate other features that needs to be included (telecoil, directional microphones and volume control). The software would list a range of specific hearing aids that meets the user's requirement to varying degrees. Appropriate selection can be made from the range displayed.

Step 6: Evaluate likely match to the prescription.

Once the hearing aid is chosen the software would indicate graphically how well the hearing aid should meet the prescription targets for the par-

ticular patient. The graphical display comprises a gain-frequency response or output frequency at one or more input levels. It is necessary to examine the match to the prescription for more than one hearing aid before taking the final decision.

Step 7: Order chosen hearing aid.

The software will usually enable the user to print out all the information necessary to order the hearing aids selected. The order can also be sent to the manufacturer electronically. For ordering custom hearing aid then the following details are also to be specified:

Battery orientation (toilet lid / swing out).
Battery size.
Telecoil and switch.
Volume control add-on cap.
Removable handle.
Microphone directionality.
Vent diameter and adjustment options.
Sound bore and earmold material.

Step 8: Retrieve patient data.

When the hearing aids have arrived and the patient is about to be fitted, one can retrieve the patient data from NOAH.

Step 9: Program the hearing aid.

The manufacturer's software via the HiPro or NOAHLink interface will make an initial adjustment of the hearing aid to approximate the prescription.

Step 10: Measure response in the patient's ear.

The response of the hearing aid should be measured with a real-ear analyzer employing a probe microphone. It would be really convenient if the results of the measurement appear on the same screen as the prescription target.

Step 11: Adjust hearing aid settings to match prescription.

Step 12: Re-measure the response in the patient's ear.

Following each adjustment of the hearing aid the response should be re-measured. After the prescription targets are achieved with sufficient accuracy the patient's reactions to all aspects of the sound quality have to be determined.

Problem solving

Most hearing aid fittings will have to be fine-tuned either electronically or physically after the patient has had a week or two to try out the hearing aids. When the patient has trouble managing the hearing aid (inserting, removing, using the controls, changing the battery) re-instruction becomes a necessity. If this problem is not solved by re-instruction then the hearing aid should be physically modified or if need be a different style be chosen. Physical modification is a must if the patient suffers discomfort from the earmold, shell or case or when the hearing aid works its way out of the ear.

Feedback oscillation has some potential solutions which include reduction of gain at selected frequencies, reducing the vent size, making a tighter earmold or shell or changing the hearing aid to one that has more effective feedback canceling algorithms.

Complaints about the patient's own voice quality are very common. The most common cause for this complaint coulld be physical blocking of the ear canal. This can be prevented by adding a vent, or by increasing the size of the existing vent which includes prefering an open-fitting hearing aid. If feedback oscillation precludes addition of vent or increasing its size then the earmold or shell should be remade with the canal stalk extending up to the bony canal. This stalk should preferably be made of a soft material. Own-voice problems are sometimes caused and cured by electronic variation of the gain-frequency response for high-level sounds.

Complaints about the tonal quality of amplified sounds are fixed by changing the balance of low-mid- and high frequency gain. The difficult part is knowing when to ask the patient to persevere with a gain-frequency response in the expectation that it will eventually become the preferred response and confer the maximum benefit to the patient.

If a patient complains about the clarity / loudness of speech or the loudness of background noise, the patient must be questioned carefully so that the acoustic characteristics of the sounds causing the problems can be identified. The aim of the clinician should be to identify whether it is the gain for low or high frequencies, and the gain for low, mid, or high levels that need to be adjusted. Only then can the appropriate hearing aid controls can be adjusted. In those cases where it is not clear which control should be adjusted, or by how much it should be adjusted, a systematic fine-tuning can be performed using either of these two methods. The first of these is paired comparisons, in which the patient is asked to choose between two amplification characteristics presented in quick succession. Multiple characteristics can be compared by arranging them in pairs. Paired comparisons can be used to adaptively fine tune a hearing aid control if the settings compared in each trial are based on the patient's preference in the preceding trial. The second general method for fine tuning relies on the patient making an absolute rating of sound quality. The best amplification characteristic is simply the characteristic that is given the highest rating by the patient. The absolute rating method can also be used to adaptively alter a chosen hearing aid control. Fine tuning is usually carried out only for patients dis-satisfied with the prescribed response, but can also be used for all patients if so desired.

Management difficulties

Users could have difficulty inserting the aid, removing it, switching it on and off, varying the volume control or changing the battery. These problems can be tackled initially by attempting further training of the user. It is also imperative to train the immediate care giver to do the task instead of the user.

Other specific solutions can also be attempted which include careful observation of the patient trying to perform the task, so that the clinician can identify precisely which part of which operation the user has difficulty performing.

Difficulty inserting an earmold or earshell:

Options for this problem include:

1. If the patient picks up the hearing aid differently each time, or in an inappropriate manner, the patient has to be taught landmarks on the hearing aid or earmold and a specific grip, and to have the procedure broken down into steps for them, which can even be written down using the patient's own words.

2. If the patient is unable to insert the helix-lock of the earmold or ITE fully into the cymba portion of the concha, it may be necessary to remove the helix lock entirely.

3. If the patient is unable to get a BTE earmold under the anti-helix, part of the earmold's conchal rim may have to be removed, turning the skeleton into a semi-skeleton for instance.

4. If the earmold/shell is a tight or tortuous fit, a lubricant may have to be applied every time the aid is inserted until the patient is more practiced / ear shape adapts to the hearing aid. Alternatively, the earmold or shell can be trimmed if feedback oscillation is not likely to result.

5. The patient may have to pull the pinna upwards and outwards with the opposite hand, while inserting the hearing aid / earmold with the ipsilateral hand.

Difficulty locating / using a control

If retraining is not successful, the hearing aid may have to be modified or replaced:

1. Volume controls on ITE / ITC hearing aids can be made more prominent with add-on caps.

2. If the patient is confused with one hearing aid with with another one, one of the controls may have to be removed so that the more important control can be operated. Some controls can be removed cleanly with electrical wire cutters.

3. The compression ratio can be increased, reducing / eliminating the need for a volume control, but potentially degrading sound quality.

4. A remote control can be provided in order to control the hearing aid controls.

Difficulty removing a hearing aid

If the patient finds it difficult to remove the hearing aid / earmold then a removal handle or line can be added or a different kind of hearing aid style can be used.

If the patient can grasp the hearing aid but not able to remove it, and cannot be trained to use an

appropriate twisting motion, parts of the earmold or shell will have to be removed or a flexible dome-type ear fitting be used instead.

Difficulty in battery changing

Options to overcome this problem could be:

1. Coloring one side of the battery slot to lessen problems with battery reversal.

2. Using a tool to open the battery compartment.

3. Using a magnetic tool to hold the battery while insertion.

4. Re-fitting with a hearing aid that has a bigger battery or a battery compartment that is easier to open / visualize.

5. Teaching the user to distinguish the positive side of the battery tactually rather than visually or vice versa. The removable tab could help with either of these approaches.

6. Fitting a hearing aid with re-chargeable battery.

Earmold / Earshell discomfort

Earmolds, earshells and BTE hearing aids can cause physical discomfort to the external ear if they apply excessive pressure at any point. The problem can be identified by asking the patient where it hurts, and also by otoscopically examining the external auditory canal for signs of inflammation. The solution to this problem is to grind away, and then polish the area of the earmold / shell that causes the problem.

For CIC aids, discomfort can also be caused by a aid that is too loose. Discomfort can occur if the patient frequently pushes the hearing aid in further than it was intended to go in an effort to retain it in the ear or to prevent feedback from occurring.

In BTE hearing aids, an incorrectly cut tubing length can create excessive pressure. Pressure spots can arise if the patient has been wearing an earmold only partially inserted. This commonly occurs if the helix lock has not been properly tucked in.

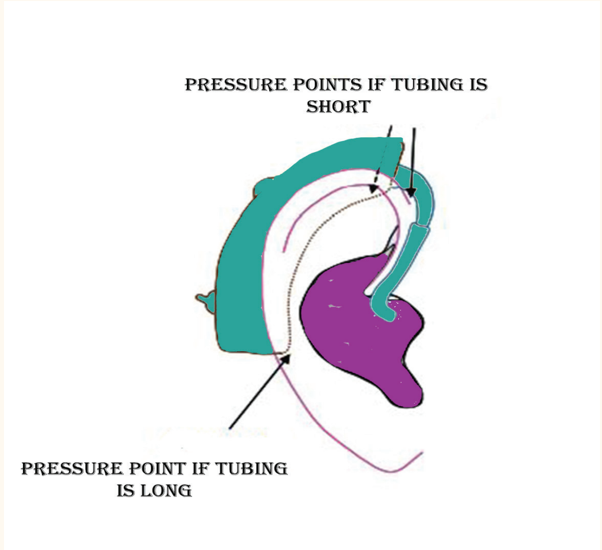


Image 1.74 showing pressure points in behind the ear model hearing aid

Generalized inflammation can be caused due to allergic reaction, but is very rare. The solution for this problem could be to re-make the earmold/shell with a different material, or to coat the mold/shell with a material that the patient is

not allergic to. Another problem could occur if the patient attempts to use the hearing aid made for his/her other ear or other person's hearing aid.

Poor earmold/earshell retention

Hearing aids (CIC / ITC styles) can sometimes fall out of the ear. Movement of the patient's jaw while speaking, yawning or chewing can move the ear canal walls sufficiently to push the hearing aid out of the ear. Solutions to this problem includes:

1. Remaking the hearing aid in a style that has better retention properties. An ITC could be used instead of a CIC or a low profile ITE can be used instead of ITC hearing aids.

2. Remaking the earmold or shell with a longer canal portion, / with a helix lock. The helix lock is in the form of a thin rim and it can be added to CIC and ITC hearing aids.

3. Remaking thee earmold or shell and taking the impression with medium viscosity material while the patient's jaw is open, so that the canal width is greater in the flexible part of the canal. The impression should extend beyond the second bend, even if the hearing aid will not be inserted this deep.

Own voice quality and occlusion

Users could complain regarding the quality of their voice as being hollow, boomy, echoes etc. Majority of users are not able to clearly describe different types of spectral emphasis and all they can say is that they do not like the way in which their voice sounds. If the user reports that other people's voices also do not osund good, then that problem should be fixed first as it could solve the self voice quality problem also. Given below are some of the reasons the patient's own voice sound abnormal to them:

The earmold/earshell is excessively blocking the ear canal:

Blocking the external canal within the cartilaginous section will allow the walls of the canal to vibrate with respect to each other, and hence could generate a high sound level in the residual part of the canal that they enclose. For low frequency sounds this could cause the SPL at the eardrum to increase by up to 30 dB relative to that which would occur for an open ear canal. Problems caused by this scenario can be solved by:

1. Increasing the area, / decreasing the length of the vent to ensure that it significantly affect its acoustic properties. An extreme example of this could be an open dome fitting which would certainly avoid any occlusion induced build up of low-frequency sound.

2. Making an earmold / shell with a canal stalk long enough to extend into the bony portion of the auditory canal. Difficulties with this solution include increased difficulty with insertion and removal and decreased comfort. The comfort can be improved by constructing the tip of the earmold out of soft material. Only drawback of this would be the life of the tip would be reduced since soft materials deteriorate faster than hard firm ones. Currently disposable soft tips are available.

3. Electronic cancellation of occlusion-generated sounds could be a thing of the future.

The hearing aid distorts when the patient is speaking

The proximity of the mouth to the ear causes the input level to the aid to be higher when the patient speaks than when other people speak, especially if the patient has a loud voice. The possibility of distortion being the cause of poor own voice quality can be tested by letting the patient hear and rate the quality of another person's loud speech using either the live voice or recorded speech as the signal. A presentation level of 80-85 dB SPL at the person's ear would be representative of own-voice levels. If distortion happens to be the problem then it is prudent to use a hearing aid that does not distort at high input levels.

Although the user is likely to prefer less gain while listening to his/her own voice than when listening to other people, this would automatically occur in an aid with WDRC. The WDRC compressor will provide lower gain for the user's own voice than it does for the voice of other persons.

Hearing aid amplifier is excessively amplifying low-frequency sounds:

Mouth radiates high frequency sounds forward more than to the side. Low frequency sounds travel around a barrier (the head) more easily than high frequency sounds. As a result the spectrum of the aid wearer's voice near his / her own ear will be more heavily weighted to low frequencies than will anybody else's voice. This bass boost occurs for everyone, but if the hearing aid is also excessively amplifying low frequencies, then the combined effect of these could cause a poor own voice quality. This problem can be diagnosed by decreasing the low frequency gain of the hearing aid for high level sounds and seeing if the problem disappears. If the haring aid uses an open fitting, then there is no point even thinking how to decrease low frequency gain because the hearing aid will have a 0 dB gain for low frequencies.

Since the frequency-dependent hearing loss affects the tonal quality of everything perceived by an unaided person, the new hearing aid user could have really forgotten what their own voice would sound like.

Feedback oscillation

This would cause the patient to report any of the following:

1. The volume control cannot be increased to the desired level without whistling occurring.

2. Whistling occurs whenever they chew, talk, wear a hat, or put their hand or a telephone near their ear.

3. The hearing aid could make a brief ringing noise whenever certain sounds occur. This is the effect of sub-oscillatory feedback or of a feedback canceler operating.

4. The hearing aid whistles when they are in a quiet place but stops when a noise occurs. This happens because WDRC causes the gain to increase in quiet places.

5. The hearing aid appears to stop working or becomes weak or distorted.

All of these problems indicate that too much sound is leaking from the ear canal to the microphone via some path. Assuming that the hearing aid is not at fault, one of the following solutions can be tried. All of them have potential disadvantages too.

Ensure that there are no excessive peaks in the real-ear aided gain curve. If so they must be damped.

It should be pointed out that the peak in the real-ear aided gain curve may be necessary to achieve the desired insertion gain curve.

If the hearing aid is vented, then the size of the vent should be reduced with a vent insert or sealing material. A change from open dome to a closed dome canal fitting could also help. This could of course exacerbate the occlusion effect. The low-frequency gain of the hearing aid may need to be reduced to offset the increase in low-frequency gain that reducing the vent diameter will have caused.

Decrease the high-frequency gain of the hearing aid, or for a multichannel nonlinear aid, decrease the high frequency compression ratio or increase the high-frequency compression threshold in the relevant channel. Some hearing aid fitting software can perform a test that identifies which channels are likely to be problematic. Disadvantage of this method is that there could be a decrease in the intelligibility or sound quality, particularly for soft sounds.

Remake or re-coat the earmold or shell so that there is less leakage between the mold/shell and the walls of the canal. Open jaw impression technique can be used if a remake is necessary. Disadvantage of this process could be the additional time and expense that it could need and the uncertainty of the outcome.

Change to a hearing aid that has a more effective feedback canceling algorithm or increase the strength off the algorithm in the current hearing aid if a choice is available for the same. Disadvantages include additional time and expense and uncertainty about effectiveness until after the alternative hearing aid has been tried.

Tonal quality

Patients could describe the quality of speech and other sounds heard in a variety of ways. Excessive high frequency amplification or insufficient low frequency amplification may be described as shrill, harsh, hissy, tinny sound. Excessive low frequency amplification or insufficient high frequency amplification could be described as dull, muffled, unclear or boomy sound. Patients are more likely to notice or adversely comment about excessive high frequency emphasis than insufficient high frequency emphasis because most patients are used to having deficient high frequency audibility when they are unaided. Solving these complaints by changing the balance of high-to low-frequency gain is complicated by three factors. One complication being an excessively peaky gain curve which could produce similar comments, even if the overall balance of low-to high-frequency gain is optimal. The solution to this problem is not to let it develop in the first place. Hearing aid fitting and verification should have included measurement of real ear gain, and this will reveal a peaky response if it existed. A peaky response should be dealt with immediately through a suitable combination of filtering and damping. Changing the damping for standard tube BTE fittings is easy because the dampers can be added to the sound tubing and can be placed in majority of earhooks. ITE, ITC and CIC receivers can also be damped, but is most easily done at the time of hearing aid manufacturing.

The second complication could be that the tonal quality could be unsatisfactory only for low-level, or only for high level sounds, or may apply across all input levels. This could easily be dealt with by appropriate questioning and the choice of which controls to adjust.

The third complication is much harder to deal with. This is the complaint of excessive shrillness

of voice. This could be because people can take months to learn to fully use high-frequency information that they have not heard previously. This is also known as the acclimatization effect.

Initially, patients could choose an amplification characteristic that gives the greatest gain for the frequencies at which they have the least loss presumably because they are most used to hearing sounds at these frequencies. Patients with a high-frequency loss are more likely to prefer high-frequency emphasis four weeks after fitting than at the initial fitting.

A compromise is to provide patients with a response that is mid-way between the response they prefer and the response that would be best for them. The aim is to enable patients to gradually get used to a new response without subjecting them to a sound quality with which they are unwilling to persevere. This could be a reasonable approach If patients wear their hearing aids every day, and have not changed their minds about what they prefer within a month then it would seem reasonable to give them the response they prefer.

The considerations are also similar for patients who are used to a peak clipping hearing aid, and are changed over to compression limiting one or for patients who are used to linear amplification and are changed over to WDRC aids. These patients favor compression limiting hearing aid after two weeks of usage.

Trainable hearing aids provide a potential solution to this dilemma. Patients can directly train the degree of high frequency emphasis.

In some hearing aids a potential solution is available. The gain in these hearing aids increases automatically and gradually in the months following hearing aid fitting. The shape of the gain-frequency response also increases. These types of hearing aids are termed as automatic adaptation hearing aids and could be programmed to gradually increase its gain for low level sounds / or its gain for high frequency sounds to provide a smooth transition towards what is best of the patient.

Noise, clarity and loudness

Adverse comments by patients will mention noise / excessive or insufficient loudness. These could be due to various reasons and each cause need a different solution. Careful history taking is essential to make sure that the patient and clinician are on the same page before taking corrective steps. It is essential to determine whether the patient is unhappy with the loudness of weak sounds, medium level sounds, or intense sounds and whether the offending sounds are sounds that the patient wants to hear including speech or other sounds. For sophisticated hearing aids that automatically change programs depending on the environment, it will be necessary to deduce which program the hearing aid is likely to be in when the adverse quality occurs.

Hearing aid is noisy in quiet places

This complaint could be an indication that internal hearing noise is being amplified sufficiently to be audible in quiet places. It may also indicate that noises in the environment are being amplified and the patient has not realized that these are noises that are present and can be heard by people with normal hearing. The clinician should listen to the aid and note should be made on whether the noise level changes when the microphone port is blocked with a finger / putty. The aim is to diagnose the noise source as being either internal or external to the hearing aid. If the problem is amplifica-

tion of low-level sounds in the environment, the noise source should be identified and the patient should be explained that normal hearing persons can also hear these sounds and that they are part of the richness of life. The patient should also be reassured that these sounds could become less noticeable as the patient becomes used to the sounds being there. The patient also should be informed that the loudness of these sounds can be decreased if the patient so desires. If complaints still persist even after prolonged use of hearing aid, it will become necessary to decrease the gain of the hearing aid for low-level sounds by raising the compression threshold or by decreasing the compression ratio.

If the problem is internal noise, then the aid should be subjected to noise measurement inside a test box. If the aid is within specification, the low level gain can be decreased by increasing the compression threshold or by introducing low-level squelch if it is available. The disadvantages are that the patient will not be able to hearing wanted low-level sounds, like soft speech. Internal noise is most likely to be heard at the frequencies where a patient has hearing thresholds that are close to normal. Internal noise in low frequencies will not be a problem for open-fit devices, as the venting effect of the open fitting attenuates low frequency noise at the output of the hearing aid. If the patient is concerned by tinnitus, then audible internal noise may be an advantage, not a disadvantage.

Soft speech in quiet places cannot be understood

The solution to this problem is to provide more gain for low-level sounds. Potential draw backs to this solution include an increased likelihood of feedback and an increased likelihood that the hearing aid will amplify sound that the person may rather not hear. It may be necessary to increase the gain in all channels, or it may be enough to increase the gain in only the low-or high-frequency channels. The frequency range requiring extra gain can be tested by having the patient comment on the audibility of speech sounds that rely on low-frequency cues (moon, boom etc) or that rely on high frequency cues (j, s etc). A low overall speech level of around 45 dB SPL should be used.

The hearing aid is sometimes too loud when noises occur

This complaint needs careful questioning. If the noises get so uncomfortable then the patient has to immediately turn the volume control down or switch off the hearing aid, the OSPL 90 of the hearing aid must be decreased. This can actually be cone by electronic variation or by increasing the damping. Increasing the damping will also decrease the gain of the hearing aid. The loudness may become uncomfortable only for sounds with significant high-frequency energy (crockery noise, paper rustling water flushing etc), it may occur only for low frequency sounds (traffic noise, door slam) or for sounds with a wide range of spectral shapes. If the problems stem from only one frequency region, it is necessary to decrease OSPL90 only in that region, assuming the hearing aid has that flexibility.

If the patient can tolerate the noise, but would rather feel better if it was not so loud so often, then a change to the input-output characteristics can improve the situation markedly. The compression ratio for input levels above 65 dB SPL should be increased. This could require the compression ratio to be increased for lower-level sounds. The output level should be maintained at a comfortable level of around 65 dB SPL for speech signals. In a multichannel hearing aid, the amount of compression could be increased in all channels.

Background noise makes it hard to understand speech

If the primary complaint is not the loudness of the noise, but rather the effect of noise on intelligibility of speech or the fatigue that is caused by trying to understand speech in the presence of noise the solution is different.

If the offending noise has a spectrum that is markedly different from that of speech the adaptive noise reduction needs to be enabled, if already enabled then its strength should be increased. Most hearing aids have adaptive noise reduction, but if the hearing aid does not have it, then the compression / gain parameters can be changed. If the offending noise is low-frequency weighted (traffic noise), increasing the amount of low frequency compression or adding a low frequency cut could help. If the noise is high frequency weighted compared to speech (crocker noise) increasing the amount of high frequency compression to achieve a TILL response, or a simple high frequency cut will help.

If the noise has a spectrum similar to that of speech, which is usually the case, and the speech has a satisfactory loudness and tone quality, then intelligibility can be improved by using an effective directional microphone or by using a remote microphone with a wireless transmission system.

Persons in the distance are easier to understand than people nearby.
The higher speech level from persons nearby may be causing excess compression or even distortion. If this is the case, then probable solutions could be to increase the maximum output or change from peak clipping to compression limiting mode.

Noise levels rise and fall intermittently.

Compression with release times from around 200 ms to 2 seconds can cause background noise levels to rise noticeably as the gain gradually rises during gaps within speech, or after brief impact sounds have forced the gain down. This audible rise and fall in the noise level is called noise pumping. The solution to this problem is to use faster compression rates.

Paired comparisons

Different response characteristics with similar perceptual effects can be selected by allowing patients to choose from two alternative responses heard in quick succession. This process is referred to as paired comparisons and patients can simply be asked which of the two conditions they prefer.

The response criterion must be made more explicit; patients can be asked which response they prefer on the basis of intelligibility, comfort, naturalness, pleasantness, minimizing annoyance of any noise present or just about any attribute of sound.

Time pressure is so intense that clinicians will have very little time to invest in more than one criterion. If paired comparisons are used to address a specific problem, the criterion should be chosen to match that problem. If the patient complains about the comfort of sound in noisy environment, the criterion should be listening comfort. If the patient is complaining about the intelligibility of soft speech, the criterion should be intelligibility of clarity. If there is lack of clarity as to the exact nature of the problem, the patient can simply be asked to choose the preferred response, with no specific criterion being mentioned.

Another important decision is what stimulus should be played to the patients while they are

choosing their preferred response characteristics. The stimulus chosen should be appropriate to the problem being addressed. If the patient is complaining about the effects of a certain type of background noise, there is little point in doing paired comparisons using speech material in quiet. If the problem relates to the clarity of speech, then a speech stimulus has to be used. In order to administer the comparisons in the minimum time, speech should be continuously present.

If the clinician does not have access to recordings of a wide range of noises, the following set of five stimuli would allow them to access hearing aid performance in such a way to address many problems reported by patients.

1. Continuous discourse in quiet, with the ability to play it at 50, 65 and 80 dB SPL.

2. Continuous discourse in quiet by three quickly alternating talkers, speaking at levels of 55, 65, and 75 dB SPL.

3. Continuous discourse at 80 dB SPL with a background noise containing high frequency impact sounds of 80 dB SPL.

4. Continuous discourse at 80 dB SPL with a background speech babble of 70 dB SPL.

5. Continuous discourse at 80 dB SPL with a background noise dominated by low-frequency sound of 80 dB (traffic noise).

The stimuli would be very useful if the speech and noise are recorded on separate channels, so that SNRs larger or smaller than those listed above can be selected when required.

The paired comparison technique can be administered by programming the settings to be compared in to different memories of the aid. The patient can switch between memories as often as desired until he or she can say which one is preferred.

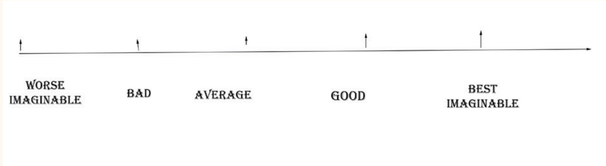


Image 1.75 showing the patient response scale for rating sound quality

Absolute rating of sound quality

One disadvantage of the paired-comparisons procedure is that it can never reveal how bad or good the sound quality is, which of the amplification schemes compared is preferred, and how much better it is than another scheme. Another disadvantage being that there are many schemes to be compared. Many comparisons of different pairs will be needed to deduce which is the best response.

Both these disadvantages can be overcome by obtaining absolute ratings of sound quality. Patients can be presented with a simple scale. These scales are termed Best imaginable, Good, Average, Bad and Worst imaginable. Patients can be instructed to provide ratings as per the scale provided.

Systematic selection by paired comparisons

In order to compare two different responses the procedure already described can be used to make the comparison. What if there are more responses that need to be compared? The best way to accomplish this depends on how many different responses that needs to be compared. Before commencing the testing process there should be a baseline response, which is the response that matches best some prescription.

Comparison to the baseline

Each response is paired, with the baseline response. Since it is not possible to have a great deal of confidence in the results of individual trial, it will be necessary to compare each of the alternatives to the baseline response several times. It is ideal that ten repetitions to be used, but realistically speaking five repetitions would be sufficient. If more than one of the responses is consistently preferred to the baseline, further comparisons will be needed to choose among them.

Round Robin

In this method each response is compared to each other, and the response that is preferred the most number of times is declared the optimal one. The Round Robin method is suitable for a small number of responses that can be used (usually four). The base line response can be added to the memory of the hearing aid.

Tournament

In this method the responses are organized in pairs and each pair is compared i.e. three times. The winners aree the ones that are preferred two or more times out of three. These winners advance to the next round, where they are gain arranged into pairs. This continues, with half of the contesting values dropping out each round until a clear winner emerges.

Irrespective of the strategy used each paired comparison judgment can be of three types:

Forced choice - Here patients have to choose a response even if they consider they can't hear any difference between the two options. People underestimate their own ability to detect small differences and to reliably choose between two similar sounds. If enough repetitions are used, the consistency of the choices indicates whether the patient is making reliable choices or whether the patient is truly guessing.

No-difference responses- In this method patients can indicate if they can't hear any difference. In this option, no points are awarded for trials where no preference is indicated. Allowing a no-difference response has been shown to increase the test-retest reliability of paired-comparison testing.

Strength of preference - In this patients will have to indicate not only which response is preferred, but also how strongly they prefer it such as slightly better, better or much better. More points are awarded for strong preferences than for weak preferences.

Some manufacturers have made it easier to administer paired-comparison testing by allowing the clinician or the patient to switch rapidly between many different amplification characteristics. These different characteristics may differ in only one amplification parameter or in many parameters. In one implementation a very large number of alternatives are represented by different positions on a

computer screen, such that characteristics differing only a little are represented near each other on the screen. The patient can use a mouse or touch screen to move around the space and indicate which parts of the screen sounds the best.

Adaptive parameter adjustment by paired comparisons

One special application of paired comparisons is determining how much a single amplification parameter should be varied. The clinician should also know by how much the compression ratio should be increased optimally. This can be achieved by a paried comparison method that can be adaptively used to find the best setting of the control. After each trial, the control setting that was not preferred is replaced with another value. The control is moved in the direction indicated by the winner of the most recent trial. The key to using this method is to choose the step size wisely. The step size should be large enough to make a perceptible difference to the quality of sound. If the step size is too small, reversals will occur randomly and the final answer will also be a random number unless a huge number of reversals are used.

Patient Education and counseling

Person with hearing impairment will benefit from education and training and counseling. These activities could be aimed at giving patients information about their hearing loss, developing skills needed to operate and care for their new hearing aids, improving listening skills, or changing patient's beliefs relating to their hearing and communication.

It would be difficult to help patients understand the variety of hearing aid styles and performance features that could be suitable for them. The benefits and cost implications will have to be presented to the patient in a simple manner. Once a person starts to use hearing aids the first time they would experience a new world of amplified sound and could benefit from guidance on how to gradually increase their range of listening experiences. The aim is to provide them with the best experiences first and to avoid having them become overwhelmed by sound. Patients need to know that their brains could take some time to adapt to hearing parts of speech and other sounds around them that they have not heard for sometime.

A major part of educating the new hearing aid user has nothing to do with hearing aids. A wide range of hearing tactics and strategies can help the hearing impaired person understand more in difficult listening situations. The first strategy requires the listener to look carefully at the talker and surroundings. The second one requires the listener to alter the communication pattern in some way. The last one and a very important one being the ability of the listener to manipulate the environment to remove or minimize sources of difficulty. Patients will definitely benefit if family members and other partners participate in education session on these topics.

It would be easy to communicate these details to the patients if it can be taught in a patient-centered, individual problem-solving method rather than as a set of rules disconnected from their everyday lives.

Patients should also be advised about protecting their residual hearing and should be made aware of where they can obtain support from peer groups ans other professionals. Hearing aids do not provide an adequate solution to all hearing problems so patients should be made aware of other assistive listening devices that could help them.

Clinicians should also be aware that different persons learn in different ways. Consequently, the same material should be taught in different ways to different persons and clinicians should develop the flexibility towards this end.

Specific aims for education and counseling related to the provision of hearing aids include:

1. Making sure th patient understands the nature of his / her hearing loss, its consequences and treatment options.

2. Helping thee patient acknowledge that he / she has a hearing loss, and working through sequential negative emotions that restrict enjoyment of life.

3. Helping the patient to overcome obstacles that discourage him/her from engaging in any form of rehabilitation.

4. Instructing and encouraging the patient in the use of hearing aids or other assistive listening devices.

5. Helping the patient acquire additional communication skills in the form off listening and communication strategies. Some of these require personal adjustment by the patient, such as increased assertiveness.

6. Providing perceptual training in understanding speech. This training can comprise analytic and synthetic speech training, in either auditory, visual or auditory-visual presentation modes.

Understanding Hearing loss

Majority of patients would like to know about their hearing capabilities and hearing loss. In order to provide accurate description of hearing loss the concepts of capability and loss should both appear in the description. The term capability indicates remaining hearing. It will help patients to understand their loss, and be able to relate it to other individuals with similar problems.

Four different aspects of hearing can be can be described to the patient. They include:

The location of their loss (the outer ear, middle ear, inner ear or the brain). This can be done with reference to a suitable wall chart or hand out.

The degree of loss (mild, moderate, severe, profound) and configuration of loss (flat, sloping etc) with reference to their audiogram and the prognosis as to how it is likely to change.

Disability / activity limitation that is to be expected should also be explained. This can be made specific by referring to the situations in which they consider they have trouble in hearing. Information about their hearing loss configuration and the influence that this has on speech discrimination may be instrumental in helping people accept that they do indeed have a hearing loss.

The handicap / participation restriction that often results in withdrawal from activities, common emotional reactions and effects on other family members should also be explained.

It should also be stressed that providing too much information could also be counter-productive and could even increase any stress they are undergoing because of the hearing deficit. Close monitoring of the patient's reactions, verbal and non-verbal is essential.

Acquiring a hearing aid

Discussion about the patient's hearing loss and its consequences automatically leads to a discussion on the various treatment options available which in most cases could be hearing aids. Assuming that the patient would benefit from the use of hearing aids other options too can be discussed with the patient.

Hearing aid style

It is probably wisest to outline the advantages of each style that apply to that patient before asking the patient which style is preferred. The patient should be allowed to physically handle each style. The patient should also be shown pictures of them when worn, or alternatively be shown what they look like when modeled in, or behind the ear of the clinician. Thin tube, or thin wire BTEs are likely to be perceived more positively relative to other styles when viewed being worn, than when viewed in the hand or on the desk. One advantage of this style of hearing aid is that if the client is ready to purchase the hearing aid at the conclusion of the hearing assessment then it can be supplied immediately, potentially saving one more

appointment for the same and thereby lowering the cost of provision.

Hearing aid technology and cost

Clinicians will have to be aware of the advantages and disadvantages of different levels of technology, and this requires knowledge of complex technologies plus the ability to assess the value of each performance feature to each patient. It is even more difficult to present it to the patient in a language that they could easily understand. The patient should have this knowledge to be able to make an informed decision about how sophisticated the hearing aids should be. It is also possible to make a table that shows the models in order of increasing sophistication, price and benefits to the patient. Majority of patients will be balancing cost against the advantages of more complex technologies.

Responsibilities and rights

Patients have a right to know how much the haring aid and associated service will cost, what warranty period covers the hearing aids, what service plans are available and what ongoing costs they will have for batteries.

Use of hearing aids

Teaching the patient how to insert a hearing aid, switch on and switch off the aid, operate the volume control, manipulate other controls if any present are essential components of patient education. Time spent on these issues will go a long way in making the use of hearing aid a comfortable one for the patient. These skills need to be mastered.

Patients need to be informed on the longevity of battery charge and its life. Patients can be provided with battery tester so that they can test the life of their batteries and replace them as and when needed. If patients are heavily reliant on hearing aids then it is imperative that they carry spare batteries along.

Since virtually all patients would like improved speech understanding in noise, and use of directional microphone is a major feature in hearing aids that could help in this aspect they should be taught two important things:

1. How to activate it by selecting the appropriate noise program. This should be chosen appropriately by the user if the hearing aid is not a fully automatic one.

2. The circumstances in which it is most effective. This knowledge is rather important as patients will be able to position themselves appropriately to achieve the best possible result. They will not be able to learn this easily if they cannot toggle between directional and omni-directional microphone settings.

The safety concerns relating to accidentally ingestion of the battery should also be covered.

Adjusting to new experiences with sound and hearing aids

When the user starts to use the hearing aid they are bombarded by an avalanche of sounds, usually in the high frequency region, that they are not used to hearing. Many users make the transition to hearing aid use easily if they build up their listening experience gradually, commencing with quiet situations and wearing the hearing aid for only a short time each day. This gradual build up is useful because patients feel encouraged by positive experiences of the aid first. Patients don't

instinctively know the situations in which hearing aids are most effective and should be guided by the clinician. Towards this end use of STEP (situations to experience and practice) forms are very useful.

Another reason for a gradual build up of listening experience is that a patient's attitude to hearing aid use may be positively affected if he/she commits to following a specified listening program.

Use of STEP form

1. The user should be explained the general principle of gradually stepping up daily listening experience both with regard to hours per day and the noisiness of the situations encountered.

2. The patient should be explained how to recognize all the sounds that he/she will be hearing.

3. It should be emphasized that this task will require certain amount of commitment and application by the patient.

4. They should be informed that the clinician is interested in their reaction to each situation, and a record needs to be maintained on how helpful the hearing aid was, and any problems that was experienced by the user.

5. It should be stressed that any situation in which the patient particularly needs to hear better is included somewhere in the list.

Typical STEP form:

1. Listening to one other person at home while you can see his/her face.

Comment -----

2. Listening to TV/radio at home.

Comment -----

3. Walking around inside the home, trying to recognize any sound that is heard.

Comment -----

4. Listening to one other person at home while the patient is not looking at their face.

Comment -----

5. Listening to music.

Comment -----

6. Listening to your own voice when you read aloud from a newspaper or book.

Comment -----

7. Conversing with two / three people in a quiet place.

Comment -----

8. Walking around outside, trying to recognize any sounds that is heard.

Comment -----

9. Shopping / talking to another person in a noisy place.

Comment -----

10. Conversing with two/three people in a noisy place.

Comment -----

11. Conversing in a large gathering or at a noisy restaurant.

Comment -----

12. Special situation.

Comment -----

The third reason for a gradual adjustment is that it reinforces to the patient that listening situations are different. If hearing aids are found to be of no use in one situation, the patient is less likely to generalize this conclusion to all other situations.

Another reason for gradual adjustment has nothing to do with sound. Earmolds and shells can cause discomfort and irritation when they are first worn, even if they fit rather well. A gradual increase in the usage hours will help the user to get accustomed to the fit. This is useful in hearing aids that extends into the bony portion of the external auditory canal.

Tailoring the patient's expectations towards the benefits offered by the hearing aid is very important. Before the user starts to actively use the hearing aid, they need to know that they will be hearing background sounds that they have not heard for so many years. They need to learn to ignore them when they carry no meaning. The clinician should explain to the patient that it might take some time to become accustomed to hearing these background sounds and to ignore them when they are not useful. This advice is very important for users of WDRC hearing aids with a low compression threshold and consequently a lot of gain for low-level sounds. Persons with normal hearing also sometimes find sounds to be annoyingly loud. Use of hearing aid augments these annoying sounds causing discomfort to the user.

Users should be warned that it could take them some months to become used to sounds provided by their hearing aids and to receive maximum benefit out of them. Formation of new neuronal pathways are necessary for this to happen. This process also goes by the name brain rewiring / acclimatization.

Hearing aid care

Patients should be trained on steps that they need to take to keep their hearing aids clean and in working condition. Patients who frequently return their custom hearing aid to the manufacturer for servicing would benefit if their hearing aids are stored overnight in a de-humidifying environment. Appropriate storage devices are available for this purpose.

Do's	Dont's
Do wipe them regularly with a tissue and occasionally with a slightly damp sponge	Don't wash them
Disconnect the BTE mold from the aid occasionally and wash the mold in warm soapy water. Tubing could take a day or so to dry out unless a hand pumper air blower is used to dry it.	Don't wear them while in shower or during swimming. Don't dry hearing aids in an oven or a microwave.
Clean the wax out of the tip, whenever it is present, with a brush, a loop or a pick or by operating or changing an in-built wax guard.	Don't insert anything more than 3 mm up the hole in the end of the aid.
Don't spray with hair spray	Store them overnight in their box or provided container.
Don't leave inside the car under sun	Battery to be removed when not in use for more than a day

Hearing strategies

Methods used by persons to increase their understanding of speech are included under this heading. Patients can use hearing strategies separately from or in conjunction with hearing aids or assistive devices. Even persons with normal hearing can use hearing strategies in difficult situations. Since hearing impaired individuals have poor speech discrimination, they would definitly need to use hearing strategies to function effectively. It is imperative that all hearing impaired persons be taught these constructive hearing strategies and should receive take home material to remind them of the important points involved.

Hearing strategies can be grouped under three categories:

Observation

Manipulating social interactions

Manipulating physical environment

Observation:

This involves observing the speaker and surroundings.

Lip-reading

A lot of information can be gained from watching peoples lips when they speak. Most people even those with normal hearing watch the speaker's lips naturally while they speak. It is hence important to instruct patients about its value as part of any rehabilitation program.

Non-verbal signals

Missing words can easily be guessed based on the topic, the speaker, facial expressions or physical surroundings. Patients will need to be encouraged to guess more often, but others will need to be encouraged to check more often.

Manipulating social interactions

All these strategies require hearing impaired persons to modify the way they interact with others. On hearing strategy used by hearing impaired people is to talk all the time so that they rarely have to listen.

Clear speaking

Some speakers are easy to understand and anyone can be easily understood when they speak clearly. Clear speech is more resistant to noise and reverberation than normal speech. Features of clear speech are:

1. Speaking rate is slower

2. Speech sound become longer

3. Vowels are fully formed

4. Stop bursts in word-final consonants are released

5, Pitch range is increased

In the case of old individuals part of the benefit from clear speech arises from the slower rate of sound production, since this allows their brains more time to process what is being heard. When there is background noise, increased speech

intensity would also improve its intelligibility because it improves Speech to noise ratio. Most hearing strategies advice that shouting could be counter-productive, but in noisy environments shouting is the only way to make oneself audible. It should be pointed out that shouting could be less of a problem in modern hearing aids with WDRC / compression limiting.

Gaining the listener's attention

Since the act of speech-reading is important, the listener should have the opportunity to speech read right from the start of an utterance. This is possible only if the listener is looking at the speaker right from the first word. Regular conversation partners could be trained to gain attention of the listener before actually talking. In adverse listening conditions this can be done by a touch, but in most circumstances it can be achieved just by saying the listener's name, then pausing, then talking.

Knowledge off he topic

This aspect makes it much easier for a person to correctly guess the words that are not heard or only partially heard. When an hearing impaired person joins a conversation with others then the first task should be to find out the topic.

Repair strategies

Breakdowns in conversation is rather common and that too when a listener has missed a key word / phrase. The listener can gain the missing information in a way that involves the minimum disruption of the ongoing conversation. These breakdowns are a problem for all persons irrespective of their hearing status. Saying 'what' repeatedly is not socially acceptable. Some of the strategies used in order to get the missed key word / phase include:

1. Repeating back the words preceding the words not heard with a question intonation accompanied by a questioning facial expression.

2. Asking a specific question that indicates what was heard and what was not.

3. Repeating back or re-phrasing what the listener thought he or she heard to confirm its correctness.

4. Asking the talker to say the last sentence or two in a different way.

5. When all else fails, asking the speaker to spell out a key word.

Giving feedback

If the patient is constantly giving feedback to the speaker then the speaker will quickly learn to speak in a way that best gets the message across without needing further intervention by the listener. Feedback comprises smiles, nods, mmm's, yes's etc.

Disclosing the hearing loss

If the patient is willing to disclose their difficulty in hearing, speakers will be prompted to make some adjustment. This could be the first big step towards accepting hearing loss in front of others.

Manipulating the environment

Lighting

Since observation of the speaker is essential for good intelligibility, good lighting is crucial in situations with adverse acoustics. The patient should

be advised to move to a postion from where he / she will be able to read the lip movements of the patient.

Positioning

The most crucial key to easy listening is position. The patient should assume a favorable position in order to hear better in challenging situations. By positioning close to the speaker, signal levels would be higher and SNR is better. Another aspect of positioning is relevant to people who have a better side and a poorer side for listening.

Minimizing noise

Noise has a disturbing effect on hearing spoken words. Solutions could be:

Turning off TV / radio down.

Closing the door

Moving to a quieter place to talk

Minimizing reverberation

Inside the home, adding soft furnishings, to a room will decrease reverberation and hence increase the intelligibility of the spoken word. Patients should choose places with such furnishings for conversations whenever possible.

Adjusting the source

When the source is an electronic appliance adjusting the tone control of the device could improve the intelligibility and naturalness.

Summary of Hearing strategies:

Watch the talker

Find out the topic

Ask the talker to speak clearly

Ask the speaker to gain your attention

Give frequent feedback

Ask specific questions

Guess meaning and repeat to confirm

Get close to talker

Get rid of noise

Discuss clear speech with others

Teaching hearing strategies

Hearing strategies can be taught in an abstract manner, but preferably taught to each patient in an individual problem solving way. One way to grab the attention of the patient is to identify a few problem situations that are important to him / her and device a list of hearing strategies that could be appropriate to each situation. This involves a candid discussion of the patient with the clinician

on the exact nature of the problem in a detailed manner.

Involving family members and friends

There are some unique advantages in involving family members and friends in the decision making process as far as the hearing aid is concerned.

Candidacy

In the first appointment patients tend to understate the hearing difficulties they are going through. Speaking to the family members / close friends could help the clinician to come to a better understanding of the problems faced by the patient. It is also emotionally satisfying for patients to know that those close to them understand the difficulties they are going through with their hearing aids. If the family member is able to appreciate the effort put by the patient in communicating to them, they can structure their visits and conversations to be shorter when the listening environment is adverse.

Hearing strategies

Most of the hearing strategies described in the previous paragraphs require active co-operation of another person.

Learning to use hearing aids

Patient's near ones / dear ones will witness the clinician instructing the patient to insert, remove and operate the hearing aid. Since they can clearly see the patient's ear they can help the patient in this task.

Auditory training

Hearing loss classically restricts patient's access to high frequency parts of speech. These patient's cortex starts to adapt itself to this scenario. This is also known as auditory plasticity / rewiring. A reversal of this process need to occur if the patient is to derive maximum benefit of hearing aids. This process is actually facilitated by systematic training in understanding speech, especially in the presence of low frequency noise.

Major goal of auditory training is to build confidence in those being trained. Auditory training can be categorized into two general types: analytic and synthetic training.

Analytic speech perception training

This training is conducted by presenting speech to the patient, requiring them to identify the sounds or to indicate whether two sounds are the same or different, and then providing feedback as to the correct answer. The aim is to help patient learn to use speech cues that should be audible to them, but for some reason they are not using them. For this type of speech training speech material is usually presented one syllable, or word at a time enabling the patient to focus on the characteristics of the sound being practiced. The material can, however be presented in whole sentences. Analytic speech perception training is also known as perceptual speech training and is used routinely used to help children develop speech perception and production skills.

Synthetic communication training

This is conducted by presenting speech to patients in a natural manner, such as by conversing with them or by having them to get, listen to a story. In synthetic training, the emphasis is on the patient understanding the message, even if the patient does not correctly perceive every sound. In this

training the listener has to synthesize (combine) any available pieces of information to correctly interpret the message. This training is also called active listening training. This implies that the listener frequently lets the talker know that the listener has understood the message. There is considerable overlap between synthetic communication training and hearing strategies. There is also some overlap between synthetic and analytic training.

Both these forms of auditory training are time consuming to perform. Analytic training has the potential to be automated.

Computer based auditory training

This helps patients to undergo auditory training in the home environment. It is also relatively easy to provide repetition and reinforcement, give immediate feedback if computer based auditory training is used. Patients with the greatest hearing impairment gain the most when computer based auditory training is used. Candidature for home-based auditory training is similar to that of hearing aids. Those with the most motivation to do it will be the most likely to complete the program and gain the benefit. Older patients are more likely to complete the program, possibly because they have more time. Premature termination of training at home can be common and clinicians should motivate the patient somehow to complete the same.

Some of the computer based auditory training programs include:

1. LACE (Listening and communication enhancement) - This comprises of tasks involving speech perception in babble and against competing talkers, time compressed speech, and closure skills (the ability for deducing missing words in sentences from context). It also provides information on hearing strategies.

2. Seeing and hearing speech - This focuses on auditory-visual speech-reading training.

3. Conversation made easy - This focuses on speech-reading training and hearing strategies.

4. Read my quips - Focuses on speech-reading training in noise using humorous sayings to maintain motivation.

Avoiding hearing aid induced hearing loss

Patients should be gently warned that wearing a hearing aid increases their risk of acquiring further hearing loss because of additional noise exposure. Patient should be advised to avoid prolonged exposure to loud noise, and to wear hearing protection in very noisy places when intelligibility is not an issue. Hearing aids will act as a form of hearing protection whenever noise level is greater than the SSPL of the hearing aids.

Assistive listening devices

During the initial phases of clinical evaluation the clinician should consider whether one or more assistive listening devices in addition to hearing aids would meet the needs of the patient. The need for assistive hearing devices should be witheld till the next follow-up appointment after the hearing aid fitting. In some patients the assisted listening device alone would be sufficient. Patients who have been provided with ALD (Assistive listening device) need significant instruction, and demonstration in its use. They need to identify situations in which they can be used.

Counseling support

This support helps patients to deal with the emotional consequences of hearing loss. If the clinician feels that the patient has emotional issues that could delay the rehabilitation process then expert help should be sought. These patients could benefit from referral to peer support groups, telephone relay services and education services.

Acts to be performed by the clinician during assessment appointment:

1. It should be determined whether the patient or someone else has initiated the appointment with the clinician.

2. Complete history should be elicited from the patient and it should include family history, etiology, work history, noise exposure, tinnitus, dizziness, asymmetry, brief medical history etc.

3. Hearing needs should be determined. This can be done by using a COSI tool (Client oriented scale of improvement designed and developed by NAL. This is a questionnaire based tool.

4. Otoscopic examination should be performed

5. Cerumen if present should be removed.

6. Otoscopic examination to be performed.

7. Hearing assessment.

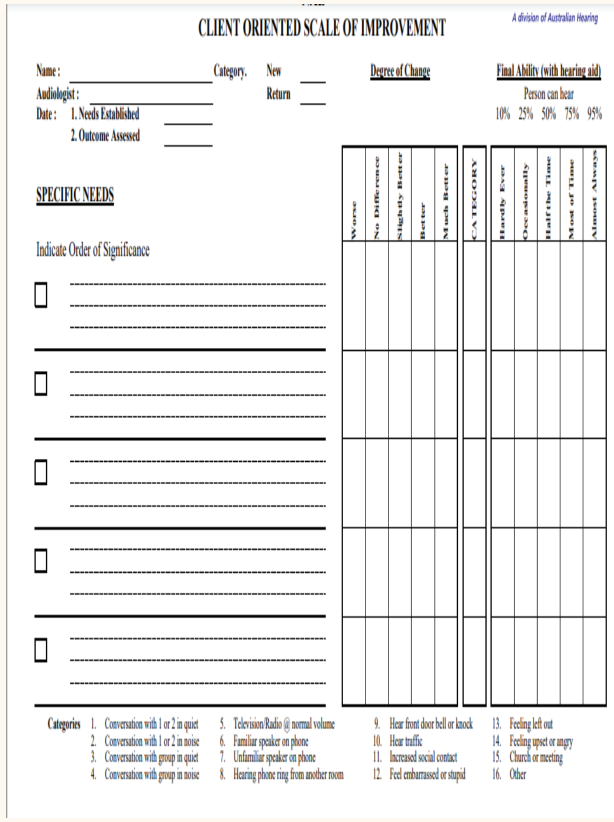
8. Expectations to b determined and modified if needed.

9. Rehabilitation options should be discussed and the patient should be informed about their advantages and limitations.

10. Likely program of fitting of hearing aid and follow-up including training options to be discussed.

11. Ear impressions to be taken if needed

12. Appropriate written report is to be prepared.


CLIENT ORIENTED SCALE OF IMPROVEMENT

A division of Australian Hearing

Name : ____ Category. New ____ Return ____

Audiologist : ____

Date : 1. Needs Established ____ 2. Outcome Assessed ____

Degree of Change

Final Ability (with hearing aid)

Person can hear

10% 25% 50% 75% 95%

SPECIFIC NEEDS

Indicate Order of Significance

Worse | No Difference | Slightly Better | Better | Much Better | CATEGORY | Hardly Ever | Occasionally | Half the Time | Most of Time | Almost Always

Categories
1. Conversation with 1 or 2 in quiet
2. Conversation with 1 or 2 in noise
3. Conversation with group in quiet
4. Conversation with group in noise
5. Television/Radio @ normal volume
6. Familiar speaker on phone
7. Unfamiliar speaker on phone
8. Hearing phone ring from another room
9. Hear front door bell or knock
10. Hear traffic
11. Increased social contact
12. Feel embarrassed or stupid
13. Feeling left out
14. Feeling upset or angry
15. Church or meeting
16. Other

Image 1.76 showing the COSI form

Fitting appointment should focus on the following tasks:

1. The hearing aid should be programed and adjusted to the patient's requirement.

2. The shell / earmolds should be modified for comfort of the patient and ease of insertion. Appropriate length of thin tube with appropriate diameter dome is chosen.

3. The hearing aid is put, adjusted the volume to suite the comfort of the patient and is left turned on.

4. The patient should be taught how to change the battery, on methods to be followed during insertion and removal of hearing aids. The patient should also be taught the differences between the left and right side hearing aids. The T switch should be demonstrated if it is present. Patient should also be trained on how to operate the volume control.

5. The quality of sound should be evaluated and fine tuning should be applied as desired.

6. Real-ear gain should be measured and the hearing aid should be adjusted to meet the prescription targets.

7. The sound quality including that of the patient's own voice should be evaluated.

8. Maximum output should be evaluated and fined tuned as needed.

9. Patient should be taught how to take care of the hearing aid which include cerumen management.

10. Use of hearing aid along with the telephone should be demonstrated to the patient.

11. Advise should be offered about the situations in which directional microphones will / will not be beneficial. This will help the patient to switch on the relevant switch that activates this feature.

12. Batteries should be provided and their approximate life and cost should be informed to the patient.

Follow up appointments

The following steps should be performed during follow up appointments.

1. The patient should be enquired about the degree of use, benefits and problems faced during hearing aid usage.

2. Usage patterns of the hearing aid can be confirmed with data logging if it is available.

3. User should be asked about the volume control setting used, adequacy of loudness, sound quality, and intrusiveness of noise.

4. Patient should be asked about the problems with own voice quality, presence of whistling and loud noises.

5. Amplification characteristics of the hearing aid can be fine tuned if indicated.

6. Patient should be asked to remove the hearing aid, change the battery, insert the hearing aid, switch it on, and adjust the volume. This is done to check the patient's ability tof manage the hearing aid.

7. Patient should be enquired about the ease of insertion and removal of hearing aid, ease of battery change etc.

8. Patient's external canal should be examined for signs of inflammation / irritation.

9. Patient should be enquired about battery consumption.

10. Patient should be enquired on how much the hearing aids have helped with their problems.

11. Patient's ability to use the telephone along with the hearing aid should he checked.

12. Need for provision of assistive listening devices is evaluated and necessary details should be provided to the patient.

13. Need for additional follow-up appointments need to be evaluated.

Formation of support groups

Formation of a group of patients using hearing aids will go a long way in helping patient rehabilitation. Some of the advantages of patient group include:

1. Some activities can be performed more effectively in a group. It also saves cost for the individual patients.

2. Some patients find participatory training with a group of similar individuals with similar problems an extremely positive experience.

3. Discussion of hearing aid problems along with the group of patients helps in finding acceptable solutions.

Patient Education and counseling

Patients and clinicians benefit from the outcomes of rehabilitation process. Systematic measurement of outcomes can help clinicians to learn which of their practices, procedures and devices are achieving the intended aims. Some measures can also help to determine how the rehabilitation program for individual patients should be structured and when it should be ended.

Outcome assessment can be based on an objective speech recognition tests or on a subjective self-report of the patient. Speech test scores could show the increase in the ability to understand speech in specific situations. Self report measures more generally reflect the patients views about the impact of rehabilitation. Patients views of their disability can be assessed both before and after the rehabilitation program. The change in score provides a measure of the effects of rehabilitation. The change in score provides a measure of the effects of rehabilitation.

There are four types of self report measures:

Standard questionnaires that directly assess benefit.

Standard questionnaires that compares disability before and after rehabilitation.

Individual questionnaires that directly assess benefit.

Individualized questionnaires that compares disability before and after rehabilitation.

Rehabilitation outcomes are likely to be affected by all aspects of rehabilitation program.

1. Clinician would want to determine if a particular rehabilitation procedure / device / entire program are more effective than others in helping his / her patients.

2. A clinician would like to ascertain whether he / she has sufficiently helped the patient. An honest answer to this question could determine whether further appointments need to be scheduled or whether a change of tactics is needed.

Outcome domains

Outcome is something that changes in life of the patient as a consequence of the services and devices provided to that patient by the clinician. Some outcomes that can be achieved include:

Decreased activity limitation - The aim of auditory rehabilitation is to enable them to hear more sounds around them and to understand speech in a range of situations. This was previously known as disability and this term was coined by WHO.

Decreased participation restriction - The aim of any auditory rehabilitation procedure should be to encourage patients not to restrict their social, occupational and recreational activities because of hearing loss. This was previously called asl handicap by WHO.

Decreased listening effort - Persons with hearing loss find it really tiring to communicate in many situations. Hearing aids are meant to decrease this effort.

Decreased emotional consequences - Hearing loss commonly leads to a range of negative emotions. Hearing aids are meant to decrease / eliminate these feelings.

Quality of life - This is a general concept and hearing aid improves the overall quality of life.

Use - Ideally patients should use these devices in every situation in which they have trouble hearing.

Satisfaction - Patients have invested time and money in participation in rehabilitation. Satisfaction of the patient and family members is important.

Tests to determune speech understanding

One of the reasons behind heaing impaired persons seeking help is to hear speech more clearly. Speech tests are a direct and objective way of measuring how much more clearly people understand speech with their hearing aids than without them. There are many speech tests that have been developed towards this end. Computer based presentation and scoring techniques enable groups of words to be scored on a phoneme basis in a reasonable time. Speech tests consequently provide a ready means to assess the benefit of hearing aids.

Despite the advantages speech tests are neither efficient nor sufficient means of demonstrating the overall benefit that hearing aids provide to a patient. Reasons for this problem include:

Dependence on measurement conditions - Amount of benefit hearing aids provide depends on the acoustic environment and the level of background noise if any. Hearing aids are least effective in noisy places where audibility is limited by background noise. Hearing aid can provide a large amount of benefit or very little benefit, depending on the target stimuli and competing sounds chosen for the test. The result cannot indicate the general benefit if the result depends on the measurement condition chosen by the clinician.

Efficiency relative to other means of measurement - Hearing aids increase speech identification ability primarily by increasing audibility. The amount by which they increase audibility depends on the speech level and spectrum, background noise level and spectrum, the patient's threshold at each frequency, and the real ear gain of the hearing aid at each frequency. Speech intelligibility index (SII) method allows us to combine them to predict aided speech intelligibility based on unaided intelligibility.

Speech tests presented at a number of levels enable a performance-intensity function to be visualized. This function enables the clinician to determine the range of speech levels over which speech scores exceed some criterion score.

Role of speech testing in evaluating benefit

1. If one can identify and simulate specific acoustic conditions, speech tests provide a clear assessment of how much the hearing aids change the person's ability to understand speech in this situation. The ability to store and quickly access a range of speech and noise materials on computers has increased the feasibility of simulating in the clinic.

2. Speech tests can provide a convincing demonstration of benefit to a patient or to a family member. This can be worthwhile if either of them is not convinced that hearing aids can provide such benefit.

3. Speech tests can demonstrate to patients and relatives the importance of visual cues to understanding.

4. Speech tests can be used to help decide whether a person should wear one or two hearing aids or in which ear a single hearing aid should be worn.

5. Speech tests can be used to predict how much

difficulty a patient will have communicating in some specified environment while wearing hearing aids.

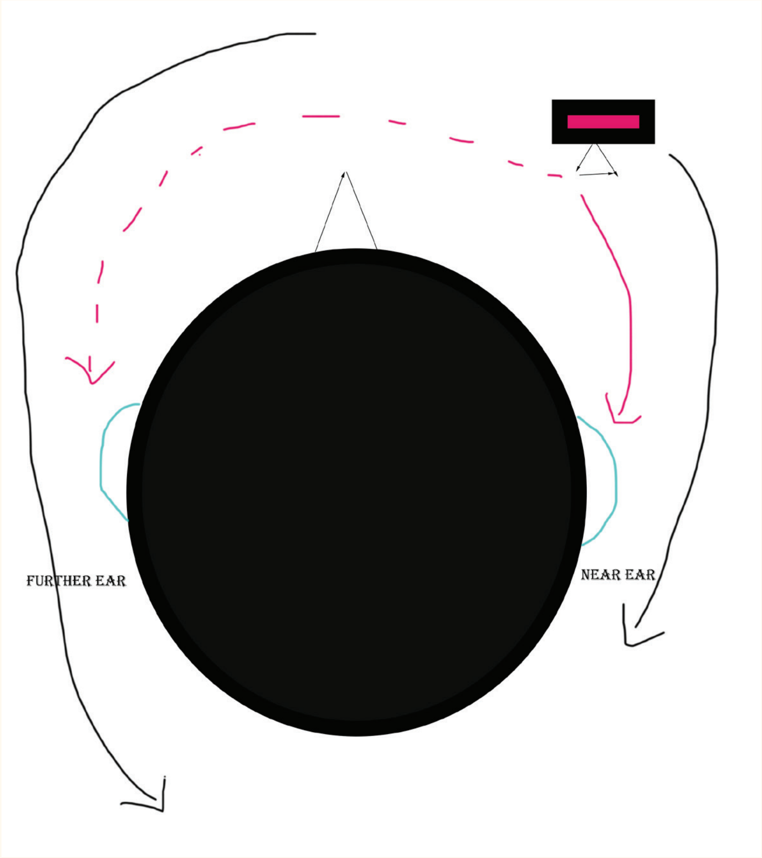


Image 1.77 showing the process of localization of sound

Binaural hearing aid fitting

Sensing sounds in two ears (binaural hearing) makes it possible for a person to locate the source of sounds and increases speech intelligibility in noisy situations. Bilateral hearing aid fitting is more important when hearing loss is severe than when it is mild or moderate. Accurate horizontal localization is possible because sounds reaching the two ears differ in level and in arrival time and hence in phase. Most persons using bilateral hearing aids can localize sounds accurately to the left or right in the horizontal plane. Vertical localization and front and back localization however are based on high frequency cues which are extremely adversely affected by hearing loss and are not significantly improved by hearing aids.

When speech and noise arrive from different directions, head diffraction causes the signal-to-noise ratio to be greater at one ear than the other. The auditory system can combine the different mixtures of speech and noise arriving at each ear to effectively remove some of the noise. This feature is known as the binaural squelch. Presenting identical sounds to two ears provides a small improvement in speech intelligibility over listening with one ear, a phenomenon known as binaural redundancy.

Wearing a second hearing aid will improve speech intelligibility in noise whenever it causes speech to become audible in the previously unaided ear. Bilateral hearing aid fitting also has other advantages which include:
Improved sound quality
Suppression of tinnitus in both ears
Hearing does not suffer even if one hearing aid breaks down

Bilateral hearing aids also have disadvantages. They cost more and are more susceptible to wind noise and are difficult for elderly people to manage.

There are obviously many advantages of listening with two ears instead of one. Using two ears instead of one enables a person to understand more when speech is heard in background noise or when there is reverberation. The ability to localise sounds is also dependent on being able to perceive sounds simultaneously in both ears. Loss of hearing in one ear will leave a person with a considerable hearing deficit in many listening situations. Similarly, when a person has a moderate to severe hearing loss in both ears but wears hearing aid in only one ear, a considerable deficit remains.

A person with a hearing aid in one ear is still able to hear many sounds in both ears and thus hears many sounds binaurally. A person with hearing aids in both ears may not hear some sounds in one ear, and it is possible that sounds heard in one ear will interfere with sounds heard in the other.

Fitting of two hearing aids is very common these days. Not everyone who receives two hearing aids wears them. There are clear evidences to prove that two hearing aids provide better performance than one. Bilateral hearing aids are a taboo for users as it involves self esteem, cost, etc.

Binaural effects in sound localization

Localization of sounds can be discussed under the following headings:

Horizontal localization - This is made possible by differences in time and intensity between the two ears. Sound will arrive at the ear closer to the source before they arrive at the ear further away from the source. The resulting difference in the arrival time at the two ears is known as the inter

Definitions:

Binaural stimulation - sounds are presented to both ears.

Monaural stimulation - Sounds are presented to one ear.

Bilateral fitting - Hearing aids are worn in both ears.

Unilateral fitting - A hearing aid is worn in one ear.

Diotic - Identical sounds are presented to both ears.

Dichotic - A different sound is presented to each ear.

aural time difference. (Image 1.77) It depends on the size of the head and the speed of sound. The inter-aural time difference is zero for frontally incident sound and increases to a maximum of about 0.7 milliseconds for sounds coming from 90 degrees with respect to the front. As the frequency increases this phase cue becomes increasingly ambiguous for sounds that do not have rapid onsets and offsets. Neural responses are highly synchronized to the sound wave form only for low frequency sounds.

Vertical localization - In the mid-sagittal plane or medial plane where there are no inter-aural clues is made possible by reflections and resonances that occur within the pinna prior to sound entering the ear canal. These reflections cause cancellations, and hence spectral peaks and notches at frequencies that depend on the elevation of the source relative to the head. The cues to localization are all above about 4 kHz because it is only in this frequency region that wavelength of the sound is small enough compared to the size of the pinna, for the necessary reflections and resonances to occur. Normal persons can detect changes in vertical angle as small as 3 degrees. Vertical localization is possible with only one ear, but performance improves slightly for listening with two ears, suggesting that brain combines the two estimates made by the information at each ear.

The additional benefit provided by the second ear may be facilitated by asymmetries between the shapes of the two pinnae that affect the pick-up of high frequency sounds.

Front-back differentiation

Externalization - This refers to the preception that a sound is originating from some point in space outside the head. In order for sounds to be externalized, the spectral shape of the sound at the two ears must have the features appropriate to the direction of the source. The transformation from SPL in the undisturbed free field to SPL in the ear canal is called the head related transfer function (HRTF). This is created by the acoustic barrier effects of the head and pinna and the direction dependent resonances of pinna.

Distance perception

Head acts as an acoustic barrier and causes a level difference between the ears. Head diffraction produces an attenuation of sound on the far side of the head and this is referred to as head shadow. Head diffraction also produces a boost on the near side of the head. Both these effects have the greatest magnitude for high frequency sounds. The resulting inter-aural level differences are much more pronounced at high frequencies.

Ears are placed almost half way between the front and back of the head. For every frontal direction there is a backward direction that results in almost the same inter-aural level and time difference cues. Pinna boosts high frequency sounds, principally in the 6-16 kHz range when they arrive from the front, but attenuates them when they arrive from the rear.

Front-back confusions are the most common type of confusions for people with normal hearing, and their likelihood increases as adults age, even for persons with good hearing. It should be realized that front-back confusions do not just apply to sounds directly in front or or behind the listener. Sounds from 30 degrees to the right of front, can be easily confused with a direction 30 degrees to the right of directly behind.

If the vertical plane is considered, there is an entire cone of directions that are confused with each other. The cone of confusable directions is referred to as the cone of confusion. Persons with normal hearing resolve directions around the cone, somewhat imperfectly by a combination of vertical localization and front-back localization.

Sound during the first few milliseconds of a signal has a strong influence on perceived direction which helps in ignoring the direction from which echoes / reflected sounds arrive. This phenomenon is known as the precedence effect, the law of the first wave-front and the Haas effect.

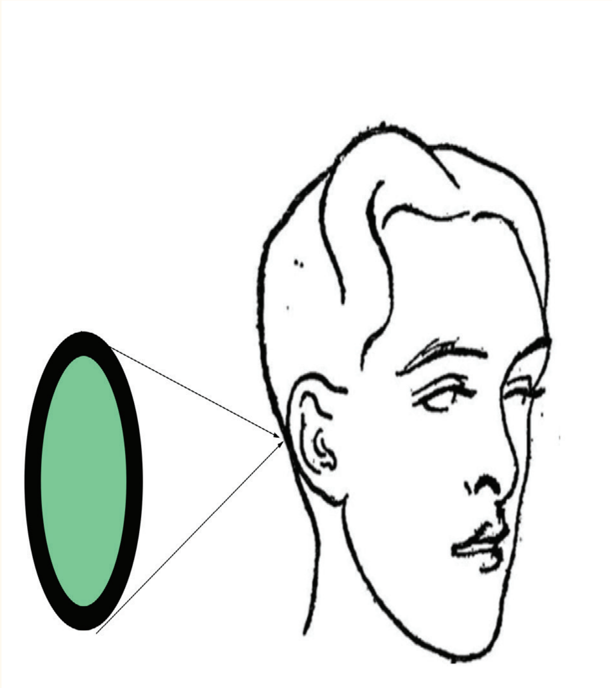


Image 1.78 showing the cone of confusion for sound localization. Any direction for which sounds have to pass through the perimeter (Black circle) to reach the ear canal are confused with each other.

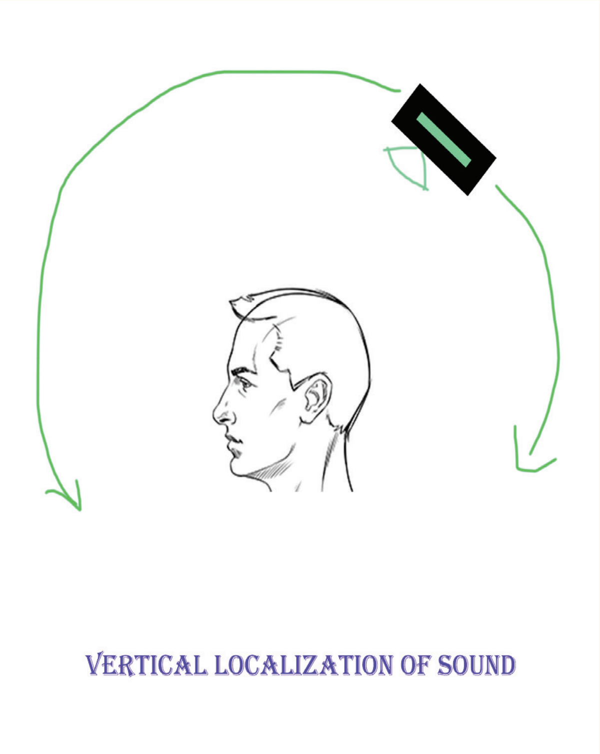


Image 1.79 showing vertical Localization of sound

Effects of hearing loss on localization

Patients with hearing loss don't spontaneously complain about poor speech localization ability. They must be specifically questioned about this aspect, particularly if they suffer severe hearing loss. When self report of patients with significant hearing loss is studied, they show significantly greater problems than normal hearing people with dynamic aspects of localization, particularly judging distance and movement.

Impaired sound localization contributes to the biggest problems in hearing loss i.e. listening to speech in noise. The difficulty in following a conversation within a group of people may be exacerbated by being unable to quickly locate the person talking, particularly when the conversation switches rapidly between speakers. The sense of localization also helps the person listening to assign a separate identity to sounds that come from different directions. Without this identification of different sound sources, multiple noises would become a confusing general background, rather than a collection of individual sounds that can be perceived and ignored if desired. Hearing impaired persons need the signal to noise ratio to be markedly higher than needed by normal-hearing people when target speech and competing sounds come from different directions. The impact of hearing loss on localization accuracy appears to be greater when there is background noise than when the target sound is presented in quiet.

In users wearing hearing aid for the first time, localization is likely to be disrupted, possibly contributing to the common experience that hearing aids seem to amplify noise more than they do to speech sounds. Poor sound localization could create a feeling of being isolated from the environment, potentially contributing to a feeling of anxiety.

Hearing loss causes front-back confusions to increase markedly, almost to the point where patients can only guess at front versus back. In real life, hearing impaired people could achieve front-back discrimination by head turning.

As low-frequency (below 1500 Hz) sensorineural hearing loss increases, horizontal localization ability gradually deteriorates. Very little deterioration occurs until low-frequency hearing sound exceeds about 50 dB HL.

In contrast to sensorineural hearing loss, conductive hearing loss causes a marked reduction in sound localization ability. As conductive hearing loss increases a greater proportion of sound that activates the cochlea is carried by bone conduc-

tion rather than by air conduction and middle ear transmission. Inter-aural attenuation is much lower for bone conducted sound than for air conducted sound.

When sound is attenuated in one ear by wearing a single ear plug, sound localization in the horizontal plane initially deteriorates greatly. In these individuals the ability to discriminate frontal from backward sources should survive the effects of occlusion because this distinction is based on the spectral shape, and not on inter-aural differences.

Hearing impairment adversely affects distance perception.

Binaural effects in sound detection and recognition

In noisy / reverberant environments people can understand speech much more accurately with two ears than with one. The ability to combine information at the two ears in order to listen to one person talking in the midst of many people talking at similar levels is often called the cocktail party effect.

There are three reasons why people can more easily understand speech with two ears than with one. The first of these arises from head diffraction effects and is a purely acoustic phenomenon. The second one is referred to as binaural squelch and relies on the brain taking advantages of differences between signals arriving at the two ears. The third one is referred to as binaural redundancy, and it also relies on the brain being able to combine signals arriving at the two ears, but does not require signals at the two ears to be different.

Head diffraction effects

An individual listening with both ears can benefit from head diffraction effects just by attending to the ear with a better SNR. The advantage arising from head diffraction is known as better ear effect. The effect on speech can be estimated by weighting the improvement in SNR at each frequency by the importance function used in speech intelligibility index. The ear nearer the speech will thus effectively have a SNR 9 dB higher than in the undisturbed field, and 17 dB higher than at the far ear.

Head diffraction will severely disadvantage a person who can hear in only one ear if the good ear is on the side of the noise and opposite on the side of the target speech. The magnitude of head diffraction effects is very large thus making speech totally understandable at one ear and totally incomprehensible at the other ear.

Head diffraction effects are purely physical in nature and in a given situation, the SNR at each frequency will be affected in the same way for a hearing impaired person as for a normal hearing person. For persons with a steeply sloping high frequency hearing loss, the benefit of head diffraction effects will be less than for normal hearing persons.

Binaural squelch in noise

It should be pointed out that simply focusing to the ear that is presented with a more favorable SNR can minimize the effects of noise. The brain and ears can do a better job than this. The auditory system can combine the signals available to each cochlea to produce an internal, central representation of the target signal that effectively has a higher SNR than is available at either ear in

isolation. This process could involve the auditory system using the noise at the ear with a poorer SNR to partially remove the noise from the ear with a more favorable SNR thereby minimizing the effects of noise.

Hearing aid issues in children

When a child is born with a hearing loss, early hearing aid provision is essential for the child to hear and speak normally. These children should be provided with hearing aid within 6 months of age. If cochlear implant is a better option then it should be performed within 12 months of age.

In order to optimally adjust the hearing aid frequency specific hearing thresholds must be determined separately for each ear. Irrespective of the type of transducer used, the small size of a baby's ear complicates interpretation of hearing threshold. This difficulty can be overcome either by expressing threshold in dB SPL in the ear canal or by expressing it as equivalent adult hearing threshold in dB HL.

Behind the ear (BTE) hearing aids are most commonly provided in conjunction with soft earmolds until the child is at least 8 years of age. The hearing aid should contain features that will enable the child to receive the best possible signal. This should ideally include an audio input socket / telecoil or internal wireless receiver so that there is some means to receive wireless transmission. The wireless device should be able to automatically attenuate the local microphone whenever the person wearing the transmitter talks.

To communicate effectively, normal hearing children learning language need a better signal to noise ratio than do adults. They also understand speech less well than adults at very low sensation levels. Hearing impaired children need more gain than adults with the same level of hearing loss. Compared to adults they do not need any more real-ear gain for high level sounds. They probably prefer more gain for low level sounds. There is a greater need for wide dynamic range compression in hearing aid for children too young to manipulate the volume control than there is for adults.

Infants have a greater need than adults for directional microphones and adaptive noise reduction systems.

To achieve a certain real-ear gain, young children need less coupler gain than do adults, because children have smaller ear canals. Age appropriate values of real-ear to coupler difference can be used to assess these values.

There are two reasons why it is important for a child to be fitted with hearing aids as early as possible. The first is to start improving the quality of life of the child and family. The second reason is that early sound deprivation has permanent effects. Neural connections in the brain that allow speech to be understood are formed based on the signals they receive from the cochlea. Neural connections can form or disappear at any stage of life, these connections are most easily formed during the early years of life. The brain's opportunity to form connections during the first two/three years of life, must not be missed if the child is to have the best possible auditory perception for the rest of his/her life. Providing habilitation early in the life maximizes language ability.

Ingredients for ensuring early and effective habilitation include:

1. Universal new born hearing screening system based on auditory brain stem response testing for all children. If OAE is used then all babies with neo-natal risk factors should receive ABR screening.

2. Seamless transfer to diagnostic testing.

3. Transfer to habilitation services, resulting in hearing aid fitting and early educational intervention before 6 months of age.

4. Implementation of one cochlear implant by 12 months of age for those infants for whom cochlear implants are likely to provide speech perception.

Any delay in the above steps would cause the child to become a deaf mute.

Binaural stimulation

Early stimulation should be binaural. Some parts of the auditory pathway combine and compare signals from the two cochleae, presumably to preform functions like localization and the binaural suppression of noise. These parts of the neural system can do their job / learn how to do their job, only if both cochleae are sending out signals. It is for this reason, binaural stimulation appears to be essential for the neuronal development that enables binaural processing of sounds.

Slight / mild hearing loss:

Situation with slight and mild bilateral hearing loss is similar to that for unilateral loss in that while several studies suggest that mild bilateral loss adversely affect educational outcomes, a few have no effect at all. There is also uncertainty as to whether fitting hearing aids improves outcomes. Fitting hearing aids to children with mild hearing loss is extremely common.

Management options for these children include:

Conventional hearing aids

FM systems

Classroom amplification

Preferential seating

Hearing aids interact with cochlear implants in these three ways:

Inadequate performance with hearing aids is an indication for cochlear implant.

In most children hearing aids should increase stimulation of the cortex prior to implantation even if speech perception is poor.

After implantation in one ear, hearing aid in the other ear provides complementary speech cues and continues stimulation of the contra-lateral pathways of the auditory system.

Auditory neuropathy spectrum disorder

There is uncertainty over the management options of children with auditory neuropathy spectrum disorder, that too during the first year of life. All device options including no device should be considered. Some reports suggest that conchlear implants is more successful than hearing aids in these children. There is a close correlation between the effectiveness of hearing aids for children with auditory neuropathy and the cortical activity elicited by speech sounds.

Older children who obtain good speech intelligibility with hearing aids appear to have cortical responses to speech sounds with normal morphology and latency, whereas children who receive minimal benefit from hearing aids do not.

If the disorder is severe and the site of the fault is within the cochlea or within the dendrites leading from inner hair cells then cochlear implant is very likely to be successful because the electrical stimulation bypasses the affected area. If the pathology is within the auditory nerve between the spiral

ganglion cells and the brainstem, cochlear implant is less likely to be successful.

Given below are some considerations that could be useful in deciding treatment options for these children:

1. If the child is unable to hear the sounds of speech unaided, then either a hearing aid or a cochlear implant is indicated as audibility is a pre-requisite for speech development.

2. Older children with robust cortical potentials evoked by sounds obtain considereable benefit in speech understanding from hearing aids. Those without cortical potentials do not benefit from hering aids. If hearing aids need to be prescribed the low gain ones should be prescribed in order to prevent noise induced hearing loss. It is argued that there is no point fitting a low gain hearing aid in which speech sounds are not heard at all. Audibility of speech sounds can be determined behaviorally in children older than 9 months of age, and by the presence of cortical evoked potentials for infants younger than this age.

3. As with other types of hearing loss, children with auditory neuropathy spectrum disorder need a better SNR than normal to understand speech. FM systems hence play an important role in these children.

4. Auditory function of some children with auditory neuropathy improves during the first year of life, particularly in low birth weight babies / hyperbilirubinemia which makes hearing aids a more conservative treatment option during this period.

Paediatritians will have to closely monitor developments in methods for managing babies with auditory neuropathy as the current knowledge of how to match treatments to these children is highly unsatisfactory.

Assessment of hearing loss in children

It is essential that hearing loss be assessed as accurately as possible within the available time frame. It is also important not to delay hearing aid fitting. The accuracy with which hearing loss can be assessed will vary with age of the child, but the techniques available now allows a reasonable asseessment to be made irrespective of the age of the child. Minimum requirements are estimated thresholds for one low frequency (500 Hz) and one high frequency (2kHz), seperately for each ear. It is better if thresholds are estimated at more frequencies. An appropriate fitting can be achieved with two reasonable accurate thresholds than with a greater number of inaccurate ones. As more audiological information is obtained during subsequent visits of the patient then the hearing aid can be fine tuned accordingly. Audiograms of children have a wider range of configuration than adults, on an average they are flatter when compared to the typical adult audiograms.

Majority of infants have approximately symmetrical hearing losses, but it would be inappropriate to assume that they all do. Hearing thresholds in the two ears measured differed by 20 dB in 10% of children. It is hence essential to obtain thresholds separately for each ear. Obtaining separate thresholds for each ear can be achieved by insert earphones, which are more comfortable than supra-aural headphones. They also reduce the need for contralateral masking because of their much greater inter-aural attenuation. These earphones can be calibrated more appropriately to fit into small ears.

Frequency specific assessment techniques per-

formed using insert earphones include:

1. Tone-burst auditory brainstem response

2. Single / multifrequency auditory steady state evoked potentials

3. Distortion product otoacoustic emissions / click evoked otoacoustic emissions.

4. Behavioral techniques such as visual reinforcement audiometry / Play audiometry

These assessments give only a generalized impression of high frequency outer cell activity rather than specific hearing thresholds.

The choice of hearing assessment technique is determined by thea ge of the child and the equipment available. In order to increase the accuracy and surety with which tresholds are estimated more than one technique should be applied. In infants preferably one behavioral measure and one electrophysiological measure and otoacoustic emissions should be included to confirm a cochlear abnormality and to help in the differentiation of sensorineural hearing loss from auditory neuropathy spectrum disorder.

Electrophysiological thresholds will initially be reported in dB nHL - the lowest stimulus level at which an electrophysiological response is reliably present, relative to the lowest stimulus level that adults with normal hearing can report hearing the same stimuli. These recorded electrophysiological thresholds must be converted into predicted behavioral thresholds for the infants by theu se of appropriate correction figures. This conversion could occur within the electrophysiological test equipment or may be done manually by the audiologist.

Small ears and calibration issues

Infant's hears and external canal are rather small. Determining hearing thresholds using insert earphones / headphones will be a problem because of calibration issues faced. Even if the transducers are calibrated so that an average adult with normal hearing has thresholds of 0 dB HL, the same cannot be applied to an infant with normal hearing.

A newborn will have an ear canal resonance nearer to 6kHz when compared to the resonance of 2.7 kHz that is seen in normal adults. If stimuli is presented via a loudspeaker, supra-aural earphones the baby's very high frequency canal resonance will make the hearing thresholds seem better than they really are at 6 kHz but worse than they really are at 3 kHz.

Baby will have a much smaller residual ear canal volume than an adult. If stimuli are presented via an insert earphone, a higher SPL will be present in the infant's ear than in the adult's ear at all frequencies. The infant will therefore appear to have less hearing loss than the adult at all frequencies, even if their middle ears and cochlea function equally effectively.

Large transducers that do not occlude the ear canal (loudspeakers / supra-aural earphones) have a low acoustic impedance so the characteristic of the ear that most affects SPL at the eardrum is the length of the ear canal. Small transducers tht fill part of the ear canal (insert earphones) have a high acoustic impedance, so the characteristic of the ear that most affects SPL at the ear drum is the volume of the residual part of the ear canal.

Hearing threhsolds in dB HL may thus appear to change during the first few years of child's life, just

because of changes in the size of the child's ears. There are two solutions to this problem. One is to express all thresholds in dB SPL in the ear canal. Expressing threshold in this way simplifies comparisons between threshold and hearing aid output expressed in the same manner. The second is to express thresholds as equivalent adult hearing level.

Auditory processing disorders

Just as in adults, children with sensorineural hearing loss have spatial processing disorder, which is a particular type of auditory processing disorder that renders them less able to attend to target sounds coming from one direction by suppresing sounds coming from other directions. Children with otitis media for more than half of their first six years are also likely to have a reduced ability to optimally combine the information present at each ear (binaural processing disorder) even after their hearing sensitivity returns to normal. Assessment of the speech perception capabilities in noise of children with spatial processing disorder, irrespective of whether they have normal hearing sensitivity or sensorineural hearing loss, will underestimate the difficulties they face unless the speech test involves presentation of speech from a direction different from that of the competing signals.

Children with auditory processing disorders and children with sensorineural hearing loss need an SNR higher than that required by their normal hearing peers. Wireless transmission from the teacher to the student is the most effective solution.

Hearing aid and earmold styles

Style of hearing aid

The appearance and size of the hearing aid is likely to be important to teen-aged children, and to parents of children of all ages. Some could choose the most brightly colored hearing aids, while others could choose more discrete ones. When worn by infants, even average sized hearing aids look very big behind tiny ears. Parents will make the final decision in the case of their children. It is essential that they first understand the likely serious consequences of choosing a hearing aid for their children. The strong link between receiving adequate signal during the first few years of life and development of speech should be emphasised. Almost all hearing impaired children are fitted with BTE hearing aids. They have the advantage over body worn hearing aids since the sound is picked up at the head level instead of being affected by clothing noise and body baffle effects, which is common if the infant is prone. They are also less likely to be soiled by food / vomit.

Body aids should be considered if for some reason BTE aids would be ineffective. This is true in children wo have additional disabilities that require their head to be supported which could cause:

Muffle sound pick up by BTE hearing aid.

Frequency bump the BTE aid.

Induce feedback oscillation in the BTE aid.

Securing hearing aids:

When children are active enough to lose their hearing aids but not old enough to make sure that they keep them, hearing aids can be secured by a Huggie aid. This is a large loop that cirlces the pinnae attached to two small loops that encircle the hearing aid. Otoclips, dino clips and Oliver clips can also be used to secure BTE hearing aids.

In the canal (ITE) hearing aids in children

There are some practical disadvantages in using ITE hearing aids in children. They include:

1. It would be difficult to fit an ITE within a small ear.

2. Small ears grow rapidly at first, and then more slowly. If an ITE is worn too early, then it will have to be replaced very frequently as the ear grows. Replacing and ITE aid is much more expensive than replacing an earmold. Ear growth stabilizes sufficiently by the age of 8-10 years to consider ITE hearing aids. With the common availability of mini-BTE hearing aids combined with thin tubing, there is little cosmetic advantage in choosing ITE devices.

2. All hearing imparied children would benefit from the use of FM or other wireless system in many circumstances. Some ITE aids do not have necessary audio input socket, wireless receiver or telecoil. This deficiency makes it impossible to combine the SNR advantages of the wireless system with the individually prescribed electroacoustics of the hearing aid.

3. There is a small risk factor with ITE aids since the shell is made of thin layer of hard plastic, breakage of hearing aid while inside the ear canal can create sharp edges that can lacerate the skin of the external canal.

4. It would be difficult for the parent / teacher to visually identify whether ITE aid is on or off and where the volume control is set.

Earmolds

Earmold designs should be carefully chosen for infants and children. Soft earmolds are more commonly used than the hard ones because of greater risk of injury from a hard mold if it is broken or pushed into the ear canal during play. Soft earmolds are less likely to cause discomfort or feedback oscillation. Soft materials ofcourse deteriorate with time, but this can be a non issue in children because anyway they need to be replaced frequently because the ear dimensions increase as the child grows.

Small size of a baby's ear can create other problems too. The angle at which the tubing protrudes from the earmold, combined with the close proximity of the earmold to the tip of the earhook can pull the hearing aid away from the surface of theh ead. The solution for this problem could be to bend the tubing more sharply, but often this cannot be done without causing a kink in the tube. A better solution could be to hollow out the centre of the conchal part of the earmold, prior to inserting the tube. The second problem created by a small ear is that it can be difficult to achieve even a 2 mm sound bore. This can be achieved by careful drilling and by terminating the sound tubing some distance from the medial tip of the earmold. It the tubing is constricted at any point, decreased high frequency gain and maximum output is likely to result. Third problem is that it could be difficult or impossible to use acoustic modifications like horns and vents. They can only be achieved in an adult sized ear. Trench vents on the outside of the earmold can be added later if needed.

Amplification prescription in children

Speech identification ability and requirements of amplification

One question that needs to be answered si that whether children / infants need amplification characteristics different from that of adults with the same degree of hearing loss. One aspect that

should be considered is related to the small size of infant's ear canals and the implications this has on the coupler gain needed. Children need less

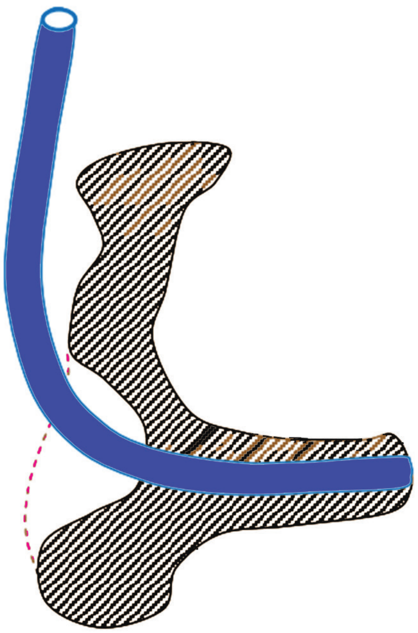


Image 1.80 showing a cross section of a hollow-concha earmold

coupler gain and OSPL90 than adults if they are to recieve the same gain and maximum output at the eardrum as adults. More difficult question that needs an answer would be whether the real-ear gain and maximum output should be different for children when compared to that of adults.

It should be stressed that when adults hear speech sound, they use their knowledge of the language to fill in any sounds that are too weak to be perceived directly. This infact occurs so naturally that they are not aware that this is happening at all. Whereas a child which is still acquiring language is less liable to fill up the missing information. The infant cannot do this at all. Even in cases of speech that contains no semantic, syntactic or lingistic context, as the case may be in nonsense syllables, normal hearing infants need the level to be about 26 dB higher than that needed by adults to discriminate between syllables with the same degree of accuracy. A 5 year old child would need levels considerably higher than older children. This would reveal that the gain should be higher for infants when compared to that of adults with the same degree of hearing loss. This difference would be greater for low level sounds than for high level ones. It is advisable not to prescribe any more high-level gain and OSPL90 for infants than adults.

According to experts children need additional high frequency gain because high frequency cues are often the most difficult for children to perceive. It should be stressed that important components of speech reside in all frequency ranges, this aspect is the cause for dilemma. Since infants discriminate speech more poorly than adults at low absolute levels, they need an SNR 7 dB higher than that needed by adults to achieve the same discrimination of nonsense syllables.

To identify familiar words and sentences, five year old children need an SNR 3-5 dB higher than that needed by older children / adults. Children will benefit from provision of wireless hearing aids in many situations, and these devices should be provided for every situation in which they can be practically used.

It is unlikely that children will benefit from more gain for high-level sounds than that given to adults, given the impact of hering aid limiting factors like loudness discomfort, and hearing loss desensitization. The close proximity that infants often have to the speaker is one of the reason against giving additional gain for high input levels. Higher gain should be provided for medium level sounds (person speaking at 65 dB SPL) provides better speech intelligibility for mid level sonds. This could ofcourse reduce the listening effort. The optimum low level gain for children (a person speaking quietly at 50 dB SPL) should be higher than that of

adults to ensure the infants / children hear better.

The inability of infants / children to alter a volume control, their decreased speech understanding at low input levels all mandate the inclusion of wide dynamic range compression hearing aids for them. WDRC aids widens the range of sound levels in the environment that after amplification lie between threshold and discomfort. With WDRC it is possible to amplify the range of sounds from soft speech to loud speech. The greatest benefit of these aids for speech perception will occur for low input levels because it increases audibility thereby increases the intelligibility of low input level speech. Since amplification on high frequency fricatives is limited children find it rather difficult to reporduce these sounds. Adaptive noise reduction would have the same advantages for children as it does in adults. It increases listening comfort, and reduces listening effort.

Hearing aid goals in children

The child should use the hearing aid consistently. Towards this goal the follwoing steps need to be taken.

Parents should be assisted to accept that their child has a hearing loss.

It must be ensured that the parents understand the close link between high quality auditory stimulation, brain development, and subsequent language development.

Parents should be explained why a specific type of hearing aid is selected.

Parents should be introduced to other parents who children are consistently using hearing aid devices.

Behavior modification plan should be designed that links use of hearing aid to some reward like book reading etc.

The child's daily routine should be identified in consultation with the parents in order to identify when aid use is practical and when it is not.

The hearing aid should be checked out for comfort when baby is assuming various positions, and it should be ensured that feedback oscillations are not present.

Parents should be provided with information about hearing aid use in the event of ear infection with suppuration.

If the child is not using the hearing aid consistently then the reason for the same should be searched for.

It should be acknowleged that achieving consistent use of hearing aid is not that easy.

Hearing aids function properly.

It must be ensured that parents can operate hearing aid controls, can insert and remove hearing aids confidently and they understand what the hearing aid does.

Parents should be provided wih their own earmolds made with 300 mm of tubing or with a stethoclip so that they can perform listening checks whild holding the hearing aid in front of them and manipulating the controls.

Parents should be demonstrated how to troubleshoot hearing aids, battery testing, usage of puffer to dry earmolds, causes of feedback oscillation. Trouble-shooting can be efficiently taught in small workshops which could train parents on various troubleshooting techniques.

The child receives high quality auditory stimulation.

Parents should be explained on the effects of noise, distance, reverberation, and head position on the quality of sound received.

If possible the clinician should demonstrate to the parent the effects of noise, distance, reverberation and hearing loss on sound quality.

It should be reinforced to the parents the need for regular, and enhanced auditory stimulation.

Use of FM sytems should be discussed with the parents including their various adjustments.

Parents understand the education options.

Basic education options should be outlined to the parent in a factual manner, especially those relating to early intervention. Parents also should know where to obtain detailed information about each option. Direct liason between audilogist and the educational institution in which the child is studying is important.

It is better to provide written information to the parent.

The child reacts to sound.

Parents should be shown comparison of amplified speech specturm (measured / calculated from real-ear gain) to hearing threhsolds.

Parents should be asked to monitor whether the child reacts to louder environmental sounds, voice of others.

CROS, Bone conduction hearing aids

In CROS (Contralateral Routing of Signals) family of hearing aids, hearing aid components on the opposite sides of the head are wirelessly linked. Basic CROS aids are most suitable for people with unilateral hearing loss.

Cros is an acronym for Contralateral routing of signals. This type of hearing aid is very useful in treating patients with unilateral severe deafness. The principle behind this type of hearing aid is that the good ear is made use of to hear sounds from the bad ear. In this type of hearing aid the microphone is placed in the bad ear to pick up sounds directed towards the bad ear.

The receiver and amplifier is fixed to the good ear. The sound picked up by the microphone is transmitted to the amplifier fixed to the good ear, it is amplified and projected to the good ear.

Advantages:

1. Since hearing is in the good ear, amplification need not be maximum

2. There is virtually no feed back

3. Directionality of sound is not lost

Modifications of CROS hearing aids:

BiCROS: This type of aid is used to treat bilateral asymmetrical hearing loss.

SteroCROS: This device is a recent innovation in cros technology. This type of hearing aid restores binaural hearing functionality of ear.

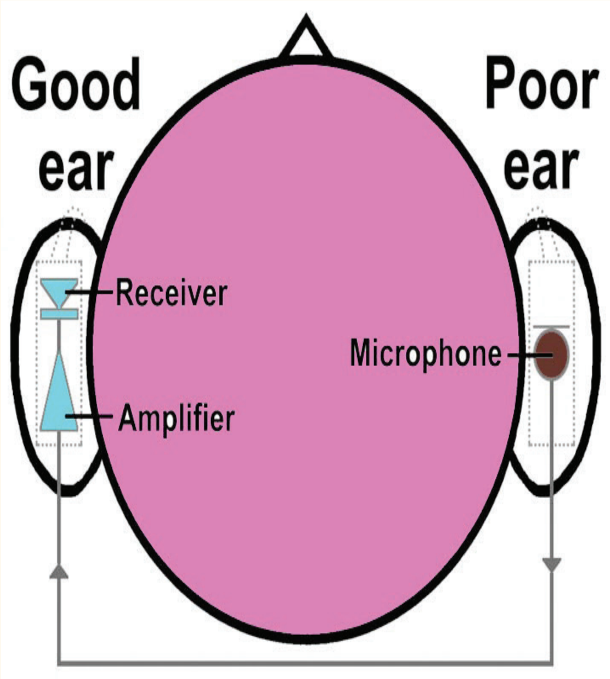


Image 1.81 showing the architecture of CROS hearing aid

In majority of cases, patients are fit either with a single hearing aid, or with two hearing aids that amplify sounds independently, or almost independently of each other. Sometimes it is better to fit people with a hearing aid system that combines components mounted at the two ears, and which requires complete audio signal to be sent from one ear to the other viaq a cable / wireless connection. This is the concept of CROS. Major disadvantage of all types of CROS aids is that a connection must be made between the two sides of the head. The most common solution is wireless transmission, fro which the major disadvantage is decreased battery life compared to self-contained hearing aids. Another solution could be a cable running around the back of the head or along the frame of a spectacle

hearing aid.

Simple CROS aids

Simple CROS aids contains a microphone mounted on the ear with the worse hearing, feeds its output to the amplifier and receiver mounted on the opposite side of the head. The seperated microphone is referred to as a satellite microphone. Signals reaching the side of the head with the poorer ear will be amplified and heard in the better ear. The receiver is coupled to the ear using an open earmold, so that unamplified sound can also directly enter the better ear. Major advantage of this system is that sounds can be heard in the ear with the better residual hearing ability no matter which direction they come from. The head acts as a baffle for high frequency sounds, boosting those sounds that come from the near side of the head and attenuating those that come from the far side. If the signal comes from one side of the listner and the predominant noise from the other side there will be a better signal to noise ratio at one ear than the other. In this scenario where signal is arriving from the poorer side, the satellite microphone of the CROS aid will pick up the relatively clear signal. The electrical connection, amplifier and the receiver will transfer the singal to the better ear.

Because of head diffraction effects, a CROS hearing aid will always improve intelligibility in noise when speech comes from the side of the poorer ear. The same diffraction effect will cause sound amplified by CROS aid to decrease intelligibility when speech comes from the side of the better ear. This disadvantage can be minimized by using no more gain than that recommended in the fitting procedure.

Another advantage of CROS aids is the microphone amd the receiver are well seperated. Signal leaking from the receiver back to the microphone is attenuated by having to pass around the head. The gain at which feedback oscillation occurs will be much higher than would be the case if the receiver and the microphone were is close proximity.

Candidacy for simple CROS aids

Patients who benefit from CROS fitting are those with unilateral hearing loss, where the loss in the poorer ear is so great that aiding it will be of no benefit. The better ear should have normal hearing or at most a mild high-frequency loss.

Patients who have near-normal low frequency hearing and moderate / severe high frequency hearing loss in one ear, combined with an unaidable loss in the other ear could also benefit for a CROS fitting. These patients would require open earmolds to avoid occlusion effects.

Adjusting and verifying CROS gain-frequency response

After the CROS aid has been pre-adjusted to approximate the coupler response, its response can be more accurately adjusted using a real-ear gain analyzer.

Step 1 - Good side response - The hearing aid is turned on. The speaker which is at 45 degrees from the front is located on the side of the good ear, The response of the ear canal of the good ear is measured. If the recorded response does not approximate the usual real-ear unaided response of an ear with no mold/ hearing aid, the mold is not sufficiently open to achieve a good CROS fitting.

Step 2 - Poor side response. The speaker is moved so that he / she is 45 degrees on the side of the poor ear. The response in the ear canal of the good ear is measured.

Step 3 - Adjust the hearing aid. If the measured re-

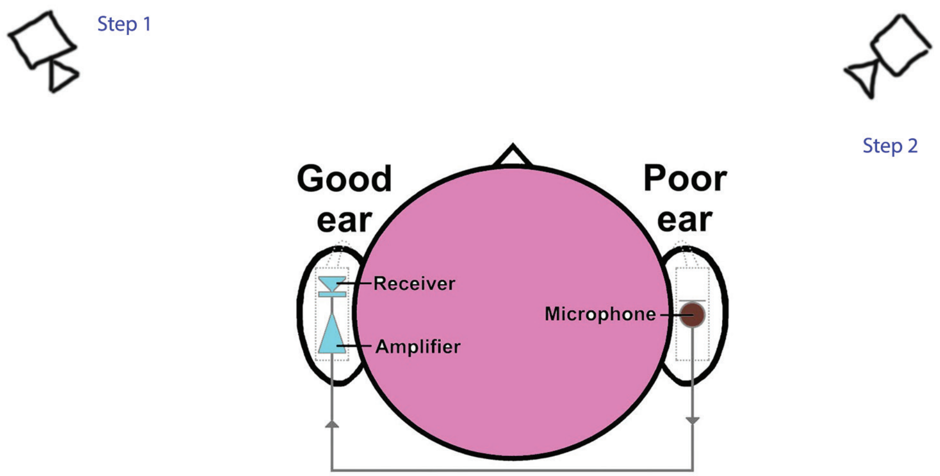


Image 1.82 showing the test set-up for verifying and adjusting gain-frequency response of a CROS hearing aid, showing two different positions of the single test speaker.

sponse in step 2 does not match with that measured in step I then the hearing aid gain is adjusted and step 2 is repeated again until the poor side response matches the good side response. If a large adjustment is to be made, it may be necessary to start again from the first step.

Step 4 - Frontal response is checked. The speaker is positioned directly in front of the patient. The real-ear gain is measured. A smoothly rising response with low-frequency gain of 0 dB and a maximum gain of 10-20 dB somewhere between 2-4 kHz should be obtained.

BICROS (Bilateral CROS) Aids

If the better ear has a hearing loss, then the patient is likely to benefit from amplification provided by a hearing aid no matter which side of the head the desired sound comes from. The only way to ensure that these signals are picked up is to have a microphone mounted to each side of the head. If each of these microphones is connected to the same amplifier and receiver then it is known as the BICROS hearing aid.

The less clear signal provided by the microphone on the side of the head further from the signal will always be added to the clearer signal from the closer microphone, thereby reducing the clarity of the clearer signal. There is of course some net benefit because the final signal-to-noise ratio will be better than that provided by a microphone on the head-shadowed side.

BICROS system works effectively for signals coming directly in front of the listener. The desired signal reaches the two microphones simultaneously and so the outputs of two microphones are added together, in phase, before being amplified. Sounds coming from other directions reach the two microphones out of phase by different degrees. The microphone outputs combine less effectively when they are added and can even cancel each other completely for particular combinations of frequency and direction. Sounds coming from directly

behind the person also arrive at the microphones in phase and are hence amplified with maximum gain.

BICROS hearing aids work as a weakly directional microphone. It also confers a minor advantage in defeating feedback.

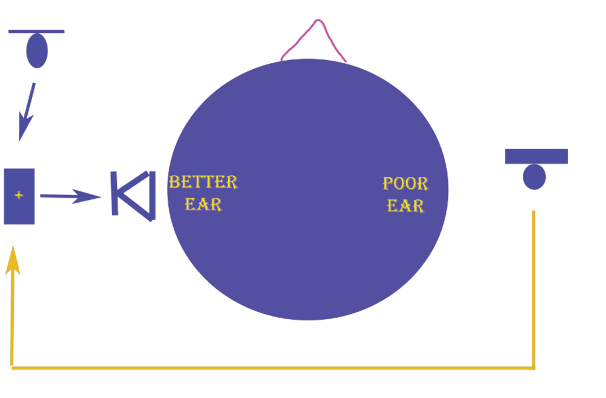


IMAGE 1.83 SHOWING BLOCK DIAGRAM OF BICROS DEVICE

Candidacy for BICROS aids

Patients would benefit from a BICROS hearing aid if they have an asymmetric bilateral hearing loss such that the poorer ear has too great a hearing loss to benefit from a hearing aid, or where amplification of the poorer ear adversely affects speech identification ability. These patients would also benefit from Bone anchored hearing aids.

Fitting procedure for BICROS hearing aids

This is a combination of fitting a conventional unilateral hearing aid and fitting a CROS hearing aid. The satellite microphone will have to have the same sensitivity as the microphone in the hearing aid.

The required gain frequency response of the hearing aids can be prescribed in the same manner as for a unilateral hearing aid, using the hearing loss of the better ear as the basis for the prescription. No allowance should be made in the prescription for binaural listening. The response should be verified with both microphones in place and the speaker located directly in front of the patient. When it is averaged across frequencies, the BICROS aids has its maximum sensitivity for frontally incident sounds.

Stereo CROS aids

In this type of hearing aid there are two separate CROS aids. The left microphone feeds the receiver on the right side, and the right microphone feeds the receiver on the left side. This solution was arrived at with the aim of achieving high gain, combined with an open fitting in both ears. The feedback path would be weak and a high gain would be possible.

Transcranial CROS aids

This is also known as power CROS or internal CROS aid. It transmits signal from one side of the head to the other using bone-conducted sound. This arrangement is useful for a person with no usable hearing in one ear, but who has to listen to sounds arriving from that same side of the head. An ITE/CIC hearing aid is fitted to the non-functioning ear. Vibrations induced on that side of the head are coupled through the bones of the head to the cochlea on the opposite side of the head. Vibrations appear to get into the skull by two paths:

The hearing aid receiver creates a relatively intense SPL in the residual hear canal volume of the dead ear, and this vibrating ear generates vibrations with-

in the temporal bone.

The hearing aid receiver vibrates the shell of the

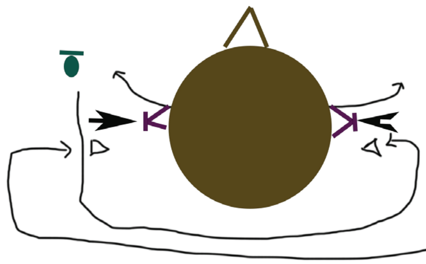


IMAGE 1.84 SHOWING STEREO CROSS AID

hearing aid, which in turn vibrates the canal wall. In order to achieve a high sensation levvel in the better ear, the hearing aid should be deeply seated within the external auditory canal.

Candidacy for trans-cranial CROS aids

If the better ear has too much sensorineural hearing loss, the trans-cranial CROS fitting will not provide enough excitation to this ear. Trans cranial CROS can improve SNR significantly only if the level of sound reaching the better cochlea via the trans cranial path is greater than the level that arrives by diffracting around the head to the eardrum of the better ear.

Bone anchored hearing aids (BAHA) are implantable hearing aids that can be used to treat hearing loss of a severe magnitude. This device works directly by stimulating the inner ear via bone conduction. It has been in use from 1977 in Europe. It got approved by the FDA only in 1996.

Indications for use of BAHA:

1. Bilateral canal atresia is an absolute indication for the use of BAHA. This is because canal atresia prevents insertion of ear mold for an air conduction aid.

2. In chronic ear infections when insertion of ear molds is a problem.

3. Unilateral deafness which cannot benefit from use of regular hearing aids.

Components of BAHA:

BAHA has 3 components. They are :

1. Titanium screw that becomes integrated with the skull bone behind the ear.

2. Titanium abutment is fitted to the titanium screw which is already integrated to the skull bone.

3. Ear level sound vibrator.

Requirements for BAHA implant:
The prime requirement for a successful BAHA implant is that the patient should have adequate bone conduction thresholds. It has been suggested that bone conduction threshold should be at least 45dB for effective functioning of BAHA. Preoperative speech audiometry should be performed in all patients before BAHA implant.

Implantation procedure:
BAHA implantation is a surgical procedure. Surgery is performed under local / general anesthesia. It is a relatively simple surgical procedure. The bone posterio-superior to the ear canal is usually of sufficient thickens over the age of three to take

an implant, and allow osseointegration. In adults mostly a single stage procedure is preferred.

In this procedure bone over the skull just postero superior to the ear canal is drilled and the titanium screw is inserted into it. Three month period is allowed to elapse for osseointegration to take place. Abutment is introduced after the osseointegration is complete. The ear level sound vibrator can be attached to the abutment.

Complications of BAHA insertion :
1. Infection

2. Crusting

3. Screw falling out

How BAHA works: It works by taking the sound from the outside and transmitting it to the inner ear through the bone. This bypasses the ear canal and the middle ear.

BAHA care:

1. Clean the area around the abutment DAILY: Washing your hair will soften any crust. Use the supplied soft cleaning brush and gently wipe the bristles against the side of the abutment, not the skin. Remove any debris around or inside the abutment. Antibacterial soap is recommended. Dry the area gently.

2. Do not allow hair to wrap itself around the abutment.

3. Do not keep hot air from a hair dryer on the abutment for a long period.

4. No hair will grow under the processor.

5. Whenever strong chemicals, such as hair dying solutions, are being applied to your hair, please cover the abutment and skin graft site with plastic to protect your skin and the abutment from the chemicals. When exposed to strong chemicals, the skin surrounding the abutment may become red, swollen, infected, or burned.

6. BAHA should be removed before undergoing MRI imaging.

7. For best benefit it should be worn throughout the day.

8. The processor will whistle when touched or when it comes into contact with other objects. Whistling can be reduced by simply repositioning the processor.

9. During windy conditions outdoors, the directional microphones may pick up wind sounds. Simply rotate the processor on the abutment 90 degrees or until the wind sound stops.

Disadvantages of bone conduction hearing aids

These hearing aids have several disadvantages over air-conduction hearing aids.

1. The transducer has to be pushed against the head with at least as much force as the peak force imparted by the vibrator. Continued use of conventional bone vibrators can cause hardened skin, permanent depressions in the skin and pain. This is caused by the relatively small area that contacts the skin, resulting in application pressure that exceeds the blood pressure inside capillaries in the skin. This causes the capillaries to collapse, depriving the tissue under the vibrator of the blood supply it needs to stay healthy. This problem can be obviated by bone vibrators with a larger contact area. These modified bone vibrators will also have a lower dis-

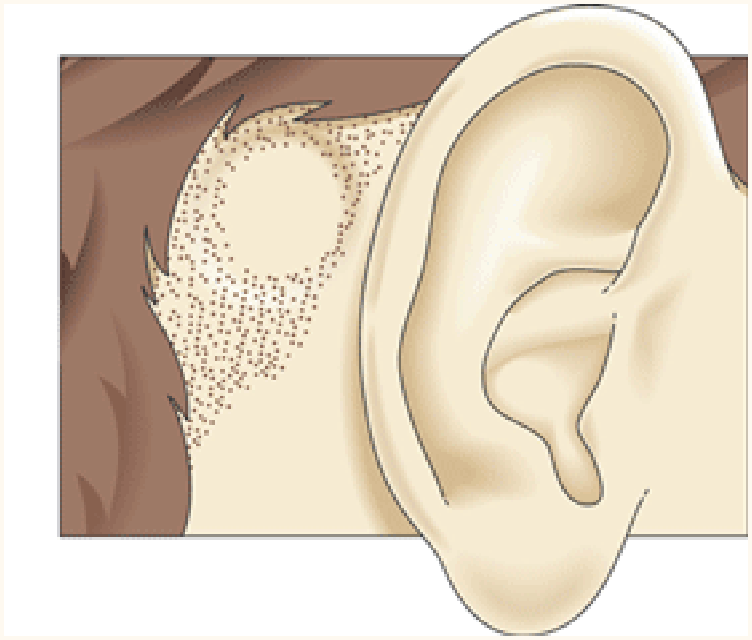


Image 1.85 showing the location of BAHA

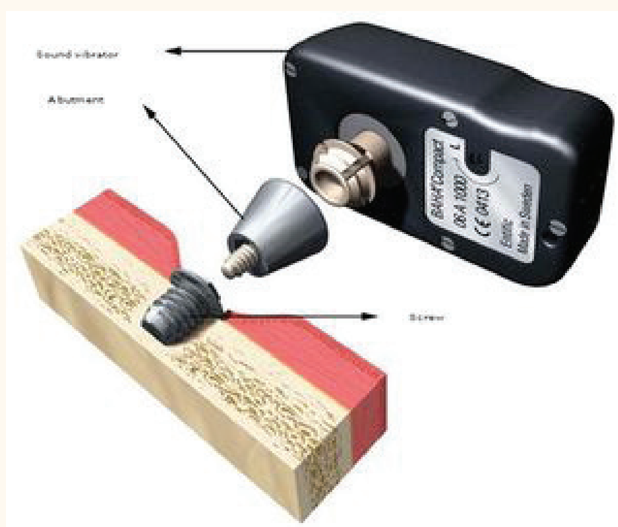


Image 1.86 showing components of BAHA

tortion and slightly higher output levels and bandwidths than conventional vibrators. The design of conventional vibrators have stayed unchanged for decades.

2. The bone vibrators are not small / discreet.

3. The inter-aural attenuation for bone-conducted sounds is considerably less than for air conducted sounds. This causes different signals to be picked from each side of the head, but it is not possible to deliver them independently to the respective cochleae. Binaural differences are smaller, but are still sufficient to allow localization and to take advantage of SNR differences between the ears made possible by head diffraction.

4. The inability to measure the output of the aid electro-acoustically makes it difficult to check the functioning of these aids as is the case for air-conduction aids.

5. The cable and plugs between the aid and the transducer can be unreliable with existing designs.

6. The attenuation provided by the skin and the limitations of the transducer make it difficult to achieve an adequate low frequency and very high frequency response. The maximum output of bone conduction aids is much less than optimal at all frequencies.

7. The transducer and headband can easily be dislodged.

The bone anchored hearing aids removes the last three disadvantages and also partially removes the first one also.

Use of BAHA for unilateral conductive/mixed hearing loss

1. Sound localization with BAHA could be slightly better than a patient with unilateral loss can achieve unaided.

2. Speech intelligibility with BAHA is likely to be better than unaided when the SNR is better on the impaired (aided) side of the head, and no different from unaided when the SNR is the same on both sides of the head.

3. Most recipients use their devices for most waking hours.

4. Most recipients subjectively report that the BAHA helps them to improve their quality of life.

Bilateral BAHA's for bilateral conductive / mixed hearing loss

Patients with bilateral conductive / mixed hearing loss can benefit from bilateral BAHAs. Although each BAHA transmits vibrations to both cochleae, the cochlea ipsilateral to the BAHA receives greater stimulation than the cochlea contra-lateral to the BAHA and receives it around 200 microseconds earlier. The amount by which stimulation of the ipsilateral cochlea exceeds that at the contra-lateral cochlea. Bilateral BAHAs can provide some degree of dichotic stimulation. The resulting inter-aural difference cues will not be as strong as for normal hearing, or for air conduction hearing aids because of significant cross stimulation relative to that which occurs of air conduction of sound. Binaural cues produced are sufficient to improve localization relative to a single BAHA.

Bilateral BAHAs provide better speech intelligibility in noise than unilateral BAHAs whenever the SNR at the unaided ear in the unilateral condition

is greater than the SNR at the aided ear.

Bilateral BAHAs also enable lower speech reception thresholds in quiet than is possible with a single BAHA. This could be partly due to each cochlea being stimulated by two BAHAs, and partly by central mechanisms benefiting from binaural redundancy.

BAHAs for single sided sensorineural hearing loss

In persons with single sided deafness a BAHA can be mounted on the side of the head with the deaf ear so that it transmits vibrations through the skull to the cochlea on the opposite side. The goal is that when the side of the head with the BAHA has the better SNR this higher quality signal will be made available to the only functioning / better functioning cochlea. The effect of BAHA on speech intelligibility varies greatly with the direction of arrival of the target speech and competing signals.

Middle ear implantable hearing aids

This is an alternative to air conduction / bone conduction hearing aids. These devices apply a mechanical vibration to the middle ear system or round window, bypassing the need for an acoustical input fro the hearing aid. Hence there is no need for an earmold. Middle ear implants can be partly implanted with the output transducer implanted but the microphone, battery, amplifier and transmitting device are worn outside of the body. If the microphone and battery are not implanted, then a signal has to be transferred from the external system to the vibrating transducer in some way. Commonly the amplified signal is sent via induction across the skin from an external coil to an internal coil and then to the internal stimulator. Other systems use a coil external to the skin to create a fluctuating magnetic field inside the middle ear which acts directly on an implanted magnet. Fully implanted hearing aids need an implanted battery that needs to be replaced every 3-10 years.

Output transducers

Middle ear implants use either electromagnetic / piezoelectric output transducers, also termed as stimulator which cause vibrations in the middle ear system / cochlea.

Floating electromagnetic transducers

This is the most common clinically applied transducer. It is composed of an electromagnetic device that relies on the inertial principle. This transducer is composed of a coil inside which a magnet is loosely suspended. When an alternating current is applied to the coil, the magnet and the coil move with respect to each other. Since the magnett has appreciable mass relative to the coil, its inertia restrains its motion, so the inertial force between the magnet and the coil causes the coil to vibrate.

If this coil is attached to one of the middle ear ossicles / round window, vibrations are transferred to the cochlea. The most common way of attaching it is to clip it onto the long process of the incus. In patients where the ossicular chain is no longer intact, the transducer can be placed in the round window niche after widening the same so that the coil presses against the round window. Alternatively it can be clipped to the stapes or presented against the footpate of the stapes thereby driving the oval window.

Although the magnet has significant mass relative to the coil, their combined 25 mg mass is sufficiently small that attachment of the complete transducer to the incus usually has only a very small effect on hearing thresholds when it is not being electrically driven once the ear has recovered from surgery.

Split electromagnetic transducers

These transducers comprise of a small permanent magnet which is vibrated by a magnetic field generated by a coil. It is otherwise physically disconnected from the coil, or any other part of the implant. When a current passes through the coil, the magnet vibrates by an amount proportional to the variations in current through the coil. This is the same principle that is used in loudspeakers. In majority of the middle ear implant systems, the magnet is firmly attached to some point in the ossicular chain, enabling vibrations to be transferred from the magnet directly to the middle ear system. Mounting points for the magnet could be tympanic membrane, incus, incudostapedial joint and the round window. The coil that drives the magnet can either be external to the ear in a hearing aid case or custom shell or in the middle ear cavity.

Anchored electromagnetic transducers

In anchored electromagnetic transducers, either the permanent magnet or the coil is attached to the bone surrounding the middle ear cavity and the and the remaining part is attached to some point in the ossicular chain. The vibrations between the magnet and coil is transferred directly to the ossicular chain. The body of the transducer is anchored within a cavity in the mastoid and transmits vibrations to the middle ear chain via a coupling link. The link can be inserted via a coupling link inserted into a hold cut by laser into the incus, or to the round window. The coupling has minimal effect on hearing thresholds when the device is inactive.

Anchored piezoelectric transducers

Piezoelectric transducers are based on a ceramic material that changes its shape when a voltage is applied to it. One end of the transducer is anchored to the bone surrounding the middle ear cavity and the free end connects to some point in the ossicular chain thus transferring sound vibrations to the cochlea. In one type of implant system, the free end terminates in a thin plate that is sandwiched between the incus and the stapes by inserting it into the incudo-stapedial joint. In another system, the vibrations are coupled to the ossicular chain by a water filled flexible tube terminating in a balloon tip. Piezoelectric transducers inherently have a wide frequency response.

All stimulators should cope with the movements of the ossicular chain in response to changes in atmospheric pressure relative to static middle ear pressure.

Microphones

Microphones for middle ear implants have been implemented in four different ways.

Externally located - These microphones for partially implanted hearing aids are located in the externally worn case, either behind the ear or against the scalp immediately lateral to the implanted electronics package. These microphones are identical to those used in conventional hearing

aids and may either be directional or omni-directional.

Subcutaneous in the scalp - Microphones in hermetically sealed cases are positioned directly beneath the skin. These cases have extra large diaphragms that are designed to minimize the loss of signal strength caused by transmission of sound through the skin.

Subcutaneous in the ear canal - These microphones are also mounted in hermetically sealed cases. They have smaller diaphragms, which decreases their sensitivity, but since they are mounted under the skin in the bony part of the external canal, their input is increased by the diffraction and resonance characteristics of the pinna and ear canal. They are known to acquire the directional properties of the normal ear.

Transducer in the middle ear

These microphones are created by locating a hermetically sealed transducer inside the middle ear cavity, positioned so that it is driven by the natural vibration of one of the structures of the middle ear. One system commercially available uses the malleus or incus motion to vibrate a hermetically sealed piezoelectric transducer. The main advantage of middle ear microphones is that they take benefit of the directional and resonance properties of pinna, concha and ear canal. The implantation surgical procedure is rather complex than placing a subcutaneous microphone.

Vibrant sound bridge

This is a partially implanted hearing aid. The externally worn audio processor contains a microphone, amplifier and battery. It inductively sends an alternating magnetic signal across the skin to a receiving coil which produces a corresponding alternating electrical signal. This is conveyed to the floating mass transducer, which is attached to the incus, or placed against the round window or less often to some other location in the middle ear system.

Carina

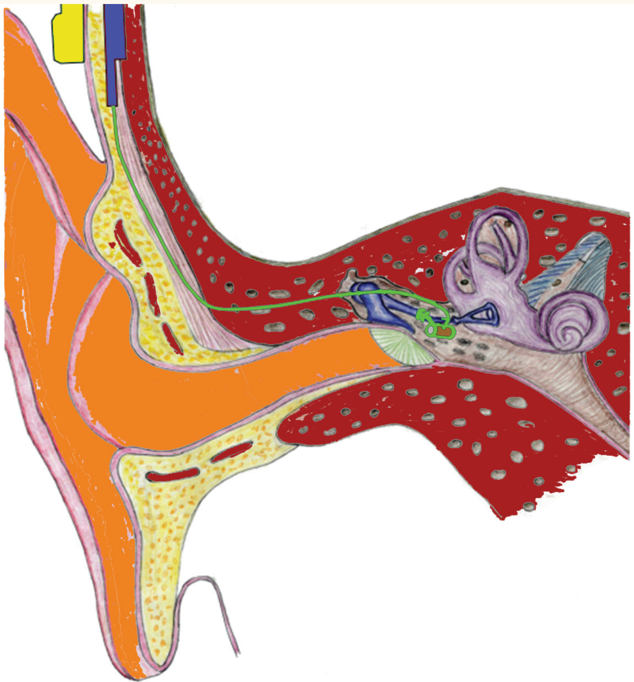
This is a fully implanted hearing aid. The microphone is mounted under the skin on the mastoid near the main package which contains the electronics and the rechargeable battery. The body of the electromagnetic stimulator referred to as the middle ear transducer is anchored in the bone of the mastoid area and connects via a partial ossicular replacement prosthesis or total ossicular replacement prosthesis to the incus, stapes, oval window or round window.

Needs a surgical procedure to insert a middle ear implant. The surgical procedure may not be always successful. Revision surgery could sometimes be needed.

Permanent alteration of taste can occur due to damage to chorda tympani nerve.

Excessively tight attachment of the floating mass transducer to incus can cause necrosis of its long process. If an implanted component is too loosely attached then it can move and create sensations unrelated to the external sound.

The potential limitation of any implanted device that includes a magnet is that it may preclude the patient from undergoing MRI scanning.



PARTIALLY IMPLANTED HEARIND AID (VIBRANT SOUND BRIDGE)
COMPRISING OF A MICROPHONE AND AMPLIFIER (YELLOW) RECEIVER AND STIMULATOR (BLUE),
OUTPUT FLOATING MASS TRANDUCER (BROWN) AND CONNECTING LEAD (GREEN).

Image 1.87 showing the components of Vibrant sound bridge

Wound infection during the implantation procedure is another potential complication that should be kept in mind.

Using Mobile Phone as assisting Device

The mobile phone revolution has ensured that almost everyone carries a mobile in his / her pocket. These phones especially android ones can be converted into an effective hearing assistive device by installing an app. A multitude of hearing augmentation applications are available in the android play store which can be downloaded and installed into android phones.

The currently available android phones are provided with excellent microphone. This microphone can be effectively used as a hearing aid microphone. The speakers available in mobile phones are of excellent quality. All the user needs to do is to wear the mobile in such a way that it would optimally capture the sound. Any person standing in front and speaking will be captured by the mobile phone and gets amplified by the amplifier present in the mobile phone. The user simply needs a head set (a wired one / blue tooth connected) in order to hear the sound.

These applications will be useful to all those persons who cannot afford to buy a new hearing aid / waiting for a new hearing aid device / have lost the hearing aid device and is waiting for a replacement.

Once installed these applications prompt the user to create an user profile. The appropriate profile can be selected by sliding up the microphone icon which could be seen in the application interface.

Applications that helps to pair the digital hearing aid to the mobile phone using blue tooth technology are also available. These applications ensure that the user is able to use the hearing aid along with the mobile phone.

The time is not far off when mobile phones embedded with digital hearing aid would reach the user as the technology for the same is already available. It is just a simple case of customization.